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Exoskeleton Power and Torque Requirements Based on Human Biomechanics

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Executive Summary

The Defense Advanced Research Projects Agency (DARPA) is funding the development of exoskeletal devices that are intended to increase the speed, strength, and endurance of soldiers in combat environments. The purpose for this work was to provide guidance for the design of the lower limbs of an exoskeletal device. In providing design guidance, the authors had two goals. The first goal was to provide estimates of the angles, torques, and powers for the ankles, knees, and hips of an exoskeleton based on data collected from humans. The second goal was to calculate the mean power required for various tasks and the total peak power needed by the lower limbs of the exoskeletal device for two "typical" infantry missions.

In order to apply human biomechanical data to design guidance for an exoskeleton, six assumptions were made:

1. The size, mass, and inertial properties of the exoskeleton will be equivalent to those of a human.

2. The exoskeleton will carry itself (including power supply) and the soldier's load.

3. The joint torques and joint powers scale linearly with mass.

4. The exoskeleton's gait will be the same as a human's gait.

5. The exoskeleton will carry a load on its back in the same way that humans carry loads on their backs.

6. The exoskeleton will move at the same speed, cover the same distance, and carry the same load as a soldier who does not have an exoskeleton.

The human biomechanical data used in this report came from studies reported in relevant journals and technical reports. These data are from studies of normal walking, walking at various speeds, walking while loads are carried, running at a moderate pace, and running at various speeds. Other activities for which joint angle, torque, and power data were obtained included stair climbing, jumping, and kneeling. The figures in Appendix A show the joint angles, joint moments, and joint powers for walking, walking while a load is carried, running, ascending stairs, descending stairs, and jumping. It is important to note that the joint power data are from calculations of the mechanical work done by the lower limbs to move the entire body, not physiological work based on oxygen consumption.

In this report, the calculation of total peak power focused on two hypothetical missions, a movement-to-contact mission and a clear-building mission. These missions represent the kind of diverse missions for which an exoskeleton might be used. Also, the fundamental tasks (walking, jogging, etc.) involved in these missions are the same tasks that occur in other infantry missions

(e.g., infiltrate and raid). The movement-to-contact mission took place over a 16-hour period, and the exoskeleton carried the soldier's sustainment load (35 kg) during most of that time. In the clear-building mission, which lasted approximately 2 hours, the exoskeleton carried the soldier's fighting load (24 kg). The soldier in these hypothetical missions was a 50th percentile male whose mass was 77 kg.

The data used to calculate total peak power for the movement-to-contact and clear-building missions came from the joint power data in Appendix A. Five simplifying assumptions about gait and the exoskeleton were made in order to calculate peak power. The assumptions were

1. Increasing the load carried has the same effect on peak power as does increasing body mass.

2. Joint powers in the frontal and transverse planes are small compared to the sagittal plane. Therefore, only sagittal plane peak power profiles were calculated.

3. Normal gait is symmetrical.

4. Peak power values for assuming and leaving kneeling and prone positions, crawling, and climbing a ladder can be approximated with the values for stair ascent.

5. Power values of other lower extremity joints, such as the metatarsophalangeal joint, are small and therefore not included in the peak power calculation.

Then, a four-step process was used to calculate the peak power required by the lower limbs of the exoskeleton during the movement-to-contact and clear-building missions.

1. Spreadsheets that listed each task in chronological order (walking, jogging, jumping, etc.), which occurred during the missions were created. The time required for each task and the load carried were also listed (see Table B-1 in Appendix B).

2. The powers required at each joint (left and right ankle, knee, and hip) were summed, and peak and average power values were identified for each task and load combination (see Figure B-1 and Table 17).

3. The peak power for each task and load combination was entered into the spreadsheet.

4. The peak power and the time required for each task were plotted for each mission (see Figures B-2 and B-3).

The accuracy of the results calculated in this report is in part a reflection of the accuracy of the original data. When biomechanical data are collected, certain assumptions, limitations, and variabilities affect the accuracy of the data[1]. The accuracy of data collected in biomechanical

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[1]For example, assumption: joint centers remain fixed throughout the range of motion; limitation: when position data are differentiated to determine velocity and acceleration, the noise in the position data becomes amplified; variability: differences in experimental methods and trial-to-trial differences in how a subject walks or runs.

studies is generally within 20% of their real values. It is likely that when these data are used to calculate kinetic variables such as joint moments and powers, compensating errors keep the overall accuracy of the results within 20% of their true values. Therefore, the power requirements calculated in this report have a tolerance of ±20%.

During most of the movement-to-contact mission, the soldier and exoskeleton walked at a natural pace, and the exoskeleton carried the sustainment load. Although the average power required for this task was 200 watts (W) ±40 W, there were peak power requirements of 500 W ±100 W. The peak powers are important because the power supply must be sized to meet the peak requirements. If the power supply cannot meet the peak requirements at the proper frequency, then the exoskeleton may, in contrast to its goal, decrease the soldier's speed, strength, and endurance. The 500 W ±100 W peaks occurred at approximately 2 Hz. This frequency coincides with ankle plantarflexion that occurs during "toe-off" in each gait cycle. When contact was made with the enemy, part of the sustainment load was dropped, and the soldier and the exoskeleton performed a variety of tasks including sprinting. The average power required during this part of the mission was 3500 W ±700 W with peaks of 5500 W ±1100 W which occurred at 4 Hz. The 4-Hz frequency coincides with ankle plantarflexion during toe-off for sprinting.

The power requirements for the clear-building mission were more variable than the power requirements for the movement-to-contact mission. This occurred because the soldier changed his speed many times as he moved to the building and then from room to room. For the clear-building mission, the average power required was 200 W ±40 W to 300 W ±60 W during much of the mission. Peak powers rose to 400 W ±80 W for walking at a natural pace and frequently to 700 W ±140 W for walking at a fast pace. There were several times when the soldier was jogging. At those times, the average power was 1000 W ±200 W, and the peak power was 2000 W ±400 W. The highest peak power requirement for this mission (approximately 3000 W ±600 W) occurred at the beginning when the soldier was running up to the building. As with the movement-to-contact mission, the peak powers must be delivered at approximately 2 to 4 Hz.

In conclusion, the data provided in this report can be used as a baseline for the initial design of an exoskeletal device. These data can be used to evaluate currently available and near-term technology to determine the feasibility of developing a practical exoskeleton. If such a device is deemed to be feasible, the joint angles, torques, and powers presented in Appendix A and Table 17 could be used to design lower limb joints and actuators, and the peak power profiles (Figures B-2 and B-3) could be used to size the power supply.

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EXOSKELETON POWER AND TORQUE REQUIREMENTS
BASED ON HUMAN BIOMECHANICS

1. Introduction

For the purposes of this report, an exoskeleton is a user-worn device that augments and enhances the wearer's speed, strength, and endurance. The exoskeleton carries its own power supply and moves synchronously with the user. Although such devices do not currently exist, efforts are under way to create them.

The Defense Advanced Research Projects Agency (DARPA) is funding a program to develop exoskeletal devices. DARPA has set requirements for very versatile machines—ones that will increase the speed, strength, and endurance of soldiers in combat environments. The requirements are set forth in a broad agency announcement (BAA) (DARPA, 2000). The BAA specifically solicits "devices and machines that accomplish one or more of the following: 1) assist pack-loaded locomotion, 2) prolong locomotive endurance, 3) increase locomotive speed, 4) augment human strength, and 5) leap extraordinary heights and/or distances." To accomplish these goals, innovative actuators and power supplies must be developed. The information in this report in intended for designers creating the actuators and power supplies for exoskeletons.

This report provides a baseline estimate of the total system power and the individual joint torques required for the lower limbs of an exoskeletal device. These estimates are based on human biomechanical data obtained from relevant journal articles and technical reports. These data are presented in the succeeding sections of this report. Because they come from a variety of sources, the data have been adapted from their original figures and tables so that they can be presented in a consistent format. The data used are from calculations of the mechanical work done by the lower limbs to move the entire body, not physiological work based on oxygen consumption.

There are two goals for this work. The first goal is to determine the angles, torques, and powers for the ankles, knees, and hips. These estimates can then be used to help in the design of joints and in selection of actuators for the exoskeleton. The second goal is to determine the total mean and peak power needed by the exoskeletal device for "typical" infantry missions. By the estimation of mean and peak power requirements, the power supply for the exoskeleton can be sized.

In the application of human biomechanical data obtained from journal articles and technical reports to the design of an exoskeleton, several assumptions are made. The first assumption is that the size, mass, and inertial properties of the exoskeleton will be equivalent to those of a human. The second assumption is that the exoskeleton carries itself (including power supply) plus the soldier's load. The third assumption is that joint torques and joint powers scale linearly

with mass. Increases in vertical ground reaction forces have been shown to be proportional to increases in load carried (Lloyd & Cooke, 2000; Tilbury-Davis & Hooper, 1999). Thus, the joint torque and power requirements for an exoskeleton are a function of the mass of the exoskeleton itself plus the load it is carrying. The fourth assumption is that the exoskeleton's gait will be the same as a human's gait. This implies that soldiers will not be required to alter their gait to use the exoskeleton. Changes in gait have been shown to increase the physiological energy expended during locomotion (McMahon, Valiant, & Frederick, 1987). The fifth assumption is that the exoskeleton will carry a load on its back in the same way that humans carry loads on their backs. This is because carrying backpack loads high and close to the back requires less energy expenditure than carrying the same loads low and farther away from the back (Obusek, Harman, Frykman, Palmer, & Bills, 1997). Also, carrying a load close to the body's center of gravity requires less energy than distributing the load to the extremities (Martin, 1985; Soule & Goldman, 1969). The sixth assumption is that a soldier with an exoskeleton will move at the same speed, cover the same distance, and carry the same load as a soldier who does not have an exoskeleton. This implies that the initial exoskeleton prototype may not allow the soldiers to perform a mission faster, but they will not become fatigued because the exoskeleton is carrying the load.

2. Typical Infantry Missions

Infantry soldiers have many different missions, and each mission is affected by such things as the enemy, terrain, weather, troops available, and time available. Soldiers participate in missions as part of a larger group—at least a squad or platoon. While it is not possible to define a "typical" infantry mission, some of the more common missions for infantry soldiers include movement to contact, react to contact, reconnaissance, infiltrate, exfiltrate, ambush, assault, raid, defend built-up area, clear building, clear wood line, clear trench line, breach obstacle, and "knock out" bunker. The fundamental tasks of soldiers during these missions include walking, jogging, running, crawling, kneeling, climbing, jumping, rolling, throwing a grenade, and firing a weapon.

The analysis in this report focuses on a hypothetical movement-to-contact mission and a hypothetical clear-building mission. These missions were chosen because they represent the kind of versatility that is expected of an exoskeleton. The hypothetical movement-to-contact mission used in this analysis takes place over a 16-hour period, and the exoskeleton carries the soldier's sustainment load during most of that time. This mission has long periods with roughly constant power requirements. However, when contact is made, there are brief periods with very high power requirements. The clear-building mission is typical of something soldiers fighting in an urban environment would do. Here, speed and strength are important assets. The clear-building mission is intense, but its duration is relatively short (approximately 2 hours). In the clear-

building mission, periods with relatively high power requirements occur more frequently than in the movement-to-contact mission. Also, large changes in the power required occur frequently. Finally, the fundamental tasks (walking, jogging, etc.) involved in these two missions are representative of the tasks that take place in the other missions listed previously (react to contact, reconnaissance, etc.).

The initial objective for the exoskeleton is for it to assist soldiers in performing their current missions. Thus, scenarios created for this analysis describe missions in the manner that soldiers who are not using exoskeletons might perform them. This is the baseline upon which system development can be founded. If the development and use of exoskeletons are successful, doctrine and tactics for various missions could change. For example, future missions by soldiers with exoskeletons would be different because the exoskeleton would carry most of the load. Therefore, the soldier with an exoskeleton would probably carry heavier loads for a longer time and at faster speeds.

3. The Application of Human Biomechanics During Locomotion to Exoskeleton Design

Figure 1 depicts a single gait cycle for the left and right legs during walking. A gait cycle is the period of time for one stride, that is, the time from one event (usually initial foot contact) to the next occurrence of the same event with the same foot. For each leg, the gait cycle can be divided into a stance phase and a swing phase. Within the stance phase, there is a period of double support: both feet on the ground. As walking speed increases, the period of double support decreases. As speed continues to increase, the period of double support can disappear, and there can be a period of "flight" when both feet are off the ground. The appearance of this flight phase is one definition of running that holds for all but a few cases.

The range of gait cycle frequencies for walking and running, based on results reported by Minetti, Ardiago, and Saibele (1994) and Fukunaga, Matsuo, Yuasa, Fujimatsu, and Asahina (1980), are shown in Table 1. The actuators for the exoskeleton need to operate at frequencies that are the same as the gait cycle frequencies—roughly 50 to 130 cycles per minute. Gait cycle frequencies also have implications for the exoskeleton's power supply. The power demands are not constant during these cycles; rather, there are peaks and valleys. For tasks such as walking, running, and stair climbing, peak power values will occur at two times during the gait cycle (once for the left limb and once for the right limb); therefore, the power supply must be able to meet these peak demands at twice the frequency of the gait cycle.

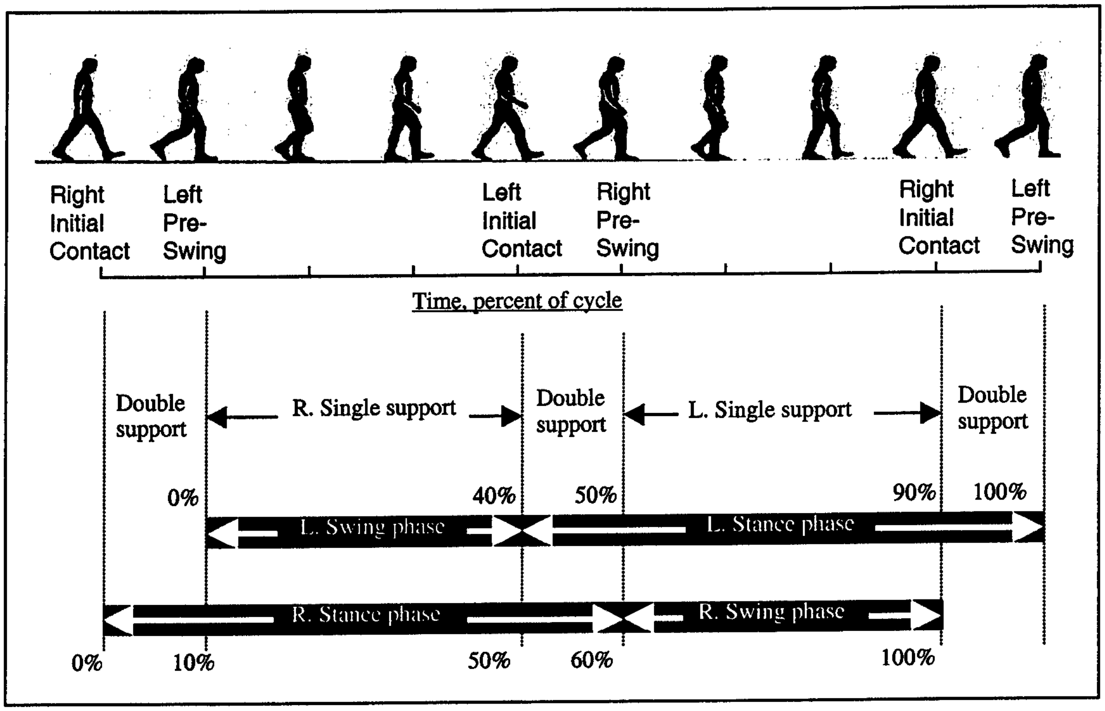

Figure 1. Gait cycle for the left and right legs during walking (adapted from a figure that appeared on page 26 of a chapter by Verne T. Inman et al., in Human Walking, edited by Rose and Gamble, published by Williams & Wilkins, Baltimore, MD; 1981, and used with permission of Lippincott Williams & Wilkins).

Table 1. Speed and gait cycle frequency

	Speed (meters/second)	Gait Cycle Frequency (strides/minute)
Walking (Minetti et al., 1994)	1.1	52
	2.4	82
Running (Minetti et al., 1994)	1.7	80
	3.3	89
Running (Fukunaga et al., 1980)	3.0	81
	6.0	96
	9.0	128

The kinematic and kinetic data collected in biomechanical studies are typically used for inverse dynamics calculations of joint forces and moments. Joint moments and joint torques are equivalent, and both terms are used throughout this report. Figure 2 shows a free body diagram of the foot at heel strike (Figure 2a) and at toe-off (Figure 2b). In inverse dynamics, the force

equilibrium and moment equilibrium equations for rigid bodies are solved to give the proximal force and proximal moment for each segment (i.e., foot, shank, and thigh). Figure 2a shows that the net joint reaction moment about the ankle is in the negative direction. The sign (positive or negative) is determined by the coordinate system used to define the data collection space. In this case, it is a right-hand coordinate system, and the subject is walking in the positive x-direction. In Figure 2b, the net joint moment is in the positive direction.

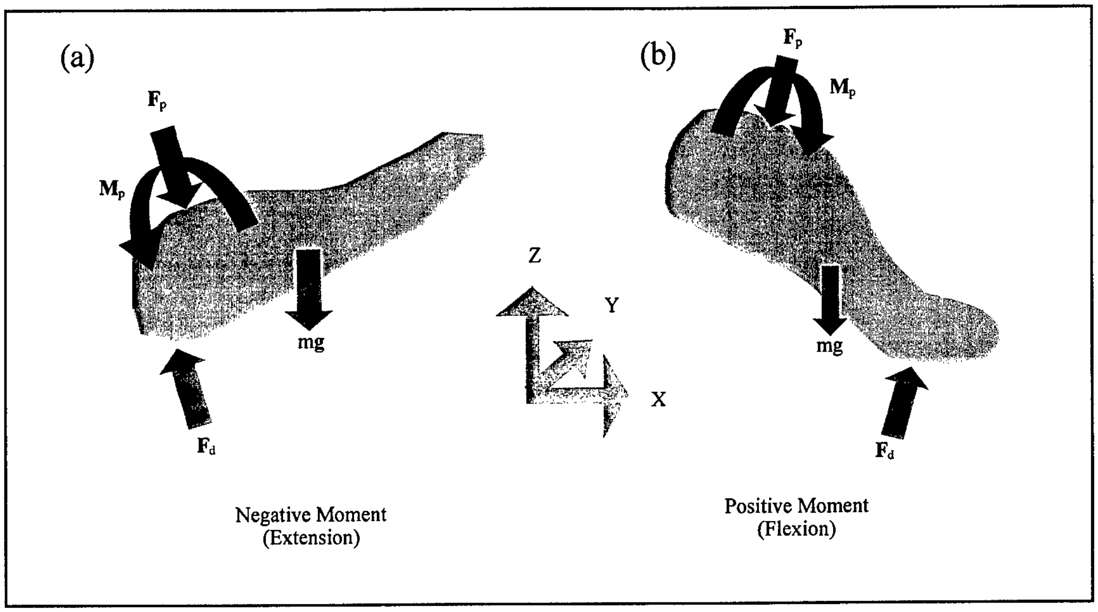

Figure 2. Free body diagram of the foot in the sagittal plane at (a) heel strike and (b) toe off. (Notation: m = mass of the segment [in this case, the foot], g = acceleration attributable to gravity, mg = weight of the segment, F_d = distal force (in this case, the force of the ground on the foot), F_p = proximal force [in this case, the force on the foot at the ankle joint], M_p = proximal moment [in this case, the moment on the foot at the ankle joint].)

Joint power (P) is the "dot product" of the moment (**M**) at the joint and the angular velocity (**ω**) of the distal segment with respect to the proximal segment (i.e., $P = \mathbf{M} \bullet \boldsymbol{\omega}$). Depending on the direction of the moment and the direction of the angular velocity, the power can be positive or negative. If the signs for the moment and angular velocity are both positive or both negative, the power is positive. If the signs for the moment and angular velocity are different, the power is negative. In the biomechanics literature, positive power is called "power generated," and negative power is called "power absorbed." The term "power absorbed" is somewhat misleading. It is important to note that power absorption is not a passive activity. In reality, the muscles are active during this time. They are doing "eccentric contractions". That is, they are generating forces while they are lengthening. For exoskeleton design, this means that the actuators need to be active and provide torques at the joints during these periods. This is particularly true of the knee after heel strike.

4. Effects of Methods and Analyses on the Accuracy of Kinematic and Kinetic Data

The following four sections are concerned with items that may have implications for an exo-skeletal device. The first of these has to do with obtaining accurate measures to determine the torque and power requirements for human locomotion because the exoskeleton is expected to provide a substantial portion of these kinetic requirements. Specifically, biomechanics researchers make several assumptions in the collection and analysis of movement data, which contribute to the accuracy of the joint kinetics. In addition to these external factors that affect data accuracy, there is also the variability in kinematics and kinetics that is associated with the neural control system, which may produce slightly different muscle activity patterns for each trial of a well-learned multiarticular movement such as locomotion. Finally, an attempt is made to determine the accuracy of kinematic and kinetic results, based on the assumptions, variability, and potential errors that exist in the collection and processing of measured and estimated data.

4.1 Assumptions

The biomechanical data presented in this report come from studies where camera systems tracked markers placed on subjects' legs (see Figure 3) as they walked, ran, or jumped on a force plate. The camera system measures the positions of the markers, and the force plate measures the ground reaction forces (GRFs) and moment. The marker position data and the force plate data are then used to calculate the kinematics and kinetics of the movement. Special computer programs use the kinematic and kinetic data to calculate joint angles, forces, moments, and powers. Also, anthropometric measures taken for each subject are used to estimate the segmental inertial properties for each subject. During this data collection process, several assumptions are generally made by the researchers to allow them to obtain these data without a prohibitive amount of effort. Indicated next are several of the main assumptions that simplify the process of collecting and analyzing the kinematic and kinetic data during locomotion and other movements.

Some of the assumptions relate to the simplified manner in which the body is modeled. The body is assumed to be comprised of a fairly small number of rigid bodies, such as the hands, forearms, upper arms, feet, etc. The entire torso is usually assumed to be a single segment or rigid body, with its length extending from the hip to the shoulder. Assuming this relatively large mass (approximately 35% of the entire body mass) as a single segment is a simplifying assumption for what is really a series of individual segments that are delineated by the articulations at the vertebrae. Related to this assumption is the fact that not every articulating joint is considered during movement analysis. As an example, the metatarsophalangeal joint of the foot is usually ignored but it may be appropriate to be considered during locomotion, particularly during running. Stefanyshyn and Nigg (2000) have shown that the maximum power absorbed at this

joint during running at 4 m/s was nearly 500 W and the energy absorbed was 27.6 J over a time period of approximately 0.14 s.

Figure 3. Subject ascending stairs with reflective markers placed on right leg.

Also related to the assumed limited number of rigid bodies and articulating joints are the degrees of freedom that exist at these particular joints. During locomotion, the sagittal plane has often been the only plane of movement that was investigated, while movements in the secondary planes (frontal and transverse) were usually ignored. Recently, however, there has been an increase in the number of studies in which researchers performed three-dimensional analyses of locomotion. These studies have shown that the kinetic variables in the frontal and transverse planes can be of significant value and should not be ignored (Eng & Winter, 1995; Glitsch & Baumann, 1997; McClay & Manal, 1999).

To obtain kinematic data, researchers place markers on the skin directly over selected anatomical landmarks that are traditionally used to represent joint centers. It is assumed that these markers remain at a fixed position with respect to the landmarks and accurately represent the joint centers for the duration of the movement. The subject's skin often slides over these landmarks, however, resulting in the markers no longer truly representing the locations of the joint centers. To add to the uncertainty of desired anatomical locations, the joint centers often do not remain at a fixed location during joint rotation. For a fairly extreme case, Smidt (1973) showed that the knee joint center moves a substantial amount when the knee joint undergoes flexion and extension.

Specifically, he found that as a subject's knee is moved between 0° and 90° of flexion, the locus of points representing the joint center at each instant forms an arc of an ellipse that is 3.2 cm in length.

Kinematic and kinetic measures are often obtained only for one side of the subject's body, assuming that bilateral symmetry exists for each variable during locomotion. This is not often the case, and research studies have shown that subjects may exhibit a substantial amount of asymmetry during movement (Sadeghi, Allard, Prince, & Labelle, 2000).

The inertial properties of the subjects are required to be known or estimated in inverse dynamic analyses. These properties include the mass, center of mass, and moment of inertia of each individual segment assumed in modeling the body. A number of methods have been used to estimate these inertial properties. The most popular method uses regression equations (Clauser, McConville, & Young, 1969) with specific anatomical dimensions used as the independent variables for these equations.

An important simplifying assumption in movement analysis is that the possible contributions of joints and ligaments to movement are ignored; that is, it is assumed that the joint kinetic values can be attributed to the muscle tendon complex only. In reality, during extreme flexion or extension at a joint, some combination of anatomical constraints and stretched ligaments may also contribute to the moment developed at that particular joint. If the actual contributions from these hard and soft tissues were considered, the portion of the joint moment value assumed to be developed by muscles spanning that joint would likely be reduced.

4.2 Limitations

Typically, the procedure to obtain linear and angular velocities and accelerations uses a motion analysis system to obtain position data, which are then differentiated by finite difference techniques, once to obtain the velocity and again to obtain acceleration data. Because of the small noise associated with the digitization process while the data are obtained and by the nature of the finite difference technique, the errors associated with the acceleration data can be quite significant. This noise can be reduced by smoothing techniques that diminish high frequency components of the noise, based on the physical justification that kinematic variables (velocity and acceleration) associated with realistic mechanical systems are band limited. This can be considered a limitation in terms of the resolution of the motion analysis equipment contributing to noise that is magnified after differentiation.

Another limitation of movement analysis studies involves how the selected sampling rate affects measures obtained from analog signals. As an example, if it were desired to obtain the time period that the foot is in contact with the ground during walking, one often would select a threshold value for the vertical force. When the force applied by the subject exceeds this threshold value, it is assumed that the subject has made contact with the ground, but in reality, contact was made at some earlier time that was between the instant when this sample was

collected and the next earlier sample. This relates to the resolution of the equipment used to measure the kinematics and kinetics of the body during movement.

For inverse dynamic analyses in which joint power is often the variable that is determined, several limitations exist. First, one cannot determine the degree of co-contraction between the agonist and antagonist muscle groups. An inverse dynamic analysis only provides net joint moment and power measures. As an example, if it is determined that the joint power generated is 60 W, that may be attributable to 60 W generated by the agonists while no power is generated by the antagonists. It could also be some combination of power in which a value greater than 60 W is generated by the agonists while the antagonists absorb an amount of power equivalent to the difference between 60 W and the power generated by the agonists. Thus, an infinite number of agonist-antagonist power combinations would be possible. This same argument that was made for joint power would also be true for the respective moments produced by the agonist and antagonist muscle groups.

Even if the moment and power values were able to be determined for a particular muscle group, it is not currently feasible for researchers to accurately determine individual muscle forces, which would be necessary to determine the individual muscle moment and power values. Forward dynamics analyses have been performed in which muscle models were used to estimate muscle kinematic (e.g., muscle and tendon velocities) and kinetic values. These studies have proved somewhat useful, but other assumptions in addition to the ones indicated before must be made. The results from these studies often deviate substantially from those found when inverse dynamics analyses are performed (Jacobs, Bobbert, & van Ingen Schenau, 1996).

4.3 Variability

Several of the assumptions and limitations indicated previously would contribute to the variability that occurs during movement, including a well-learned task such as locomotion. In addition, other factors contribute to the variability encountered in any type of unrestricted movement. One of these factors would include the location and orientation of the principal axes of each segment. Different researchers may select different anatomical landmarks to define the principal axes of the segments. This has resulted in a substantial variation in joint angles reported at any instant across studies while the kinematic trends over a locomotion cycle may be quite similar among these studies.

For two-dimensional analyses, a factor that contributes to the variability of kinetic data is the location of the center of pressure. The center of pressure location is a representation of where the centroid of the GRF is positioned on the shoe in contact with the force platform. Because of the potential difference in foot structure among individuals, the actual location on the foot on which the center of pressure acts could be slightly different among individuals wearing the same footwear. In addition, the orientation of the foot (with respect to the horizontal axis of the force platform in the direction of walking) would contribute to a researcher's correctly locating the actual center of pressure on the foot. In a two-dimensional analysis, the greater the angle

between the longitudinal axis of the foot and this force platform axis, the greater would be the error in the center of pressure location on the foot because the axis of the foot would be outside the two-dimensional (sagittal) plane in which movement is assumed. These errors, in turn, would contribute to the variability in the moment arm of the GRF acting about the ankle joint. In addition, when the GRF values are fairly large, their moment arms are quite small compared to overall dimensions of the foot, and thus, the kinetic variables would be fairly sensitive to the center-of-pressure location.

In a study that specifically investigated this issue, McCaw and DeVita (1995) determined the variation in joint torques during the stance phase of gait if the anterior-posterior location of the center of pressure was in error by 1.0 cm. They found that this center of pressure error resulted in an average change of 14% in the joint moments at the hip, knee, and ankle, and they concluded that published joint moment data for gait may be in error by 7% to 14% solely because of an inaccurate center-of-pressure location.

In general, a certain amount of variability can also be attributed to the difference in body structure among individuals. While inertial property estimates may be quite accurate for an individual, it is not uncommon for these estimates to be generally in error by as much as 10% of the actual value. These errors would be carried through the calculations in the determination of kinetic measures in an inverse dynamics analysis. As an example, Nagano, Gerritsen, and Fukashiro (2000) simulated errors of 10% of segment length for the joint center and segment center of mass locations and determined that the work output obtained was in error by 20% and 6%, respectively. Challis and Kerwin (1996) determined errors in the joint moment for a loaded elbow flexion movement when kinematic, segmental inertial property, and joint center values were varied by amounts consistent with measurement uncertainties for these respective variables. They found that varying the kinematics resulted in the largest changes in elbow joint moment, producing a value that was 21% different from the peak joint moment.

In a fairly comprehensive study of joint moment variability during walking, Winter (1984) obtained within- and between-subjects measures of variability. He provided plots of joint positions, GRFs, and joint moments and determined a coefficient of variation (CV) based on the root mean square of the standard deviation. In addition, he determined a support moment that is the algebraic sum of the moments at the ankle, knee, and hip joints and represents the net extensor moment during walking. For nine trials for a single subject, the CV for the joint angles, GRFs, and ankle joint and support moments were relatively small, while the knee and hip joint moments were approximately three times as large. When similar data were obtained for 14 to 16 subjects walking at fast, normal, and slow speeds, the same trends in variability across the kinematic and kinetic variables were obtained although the CV values were approximately two (fast speed) to three (slow speed) times larger than those obtained for the single-subject analysis. For this between-subject phase of the study, the kinetic variables were normalized to body mass to account for weight differences among the subjects. These results were not surprising, given the likely greater variability in speed across trials for a particular cadence and the variation in leg

length (and thus, stride length) among subjects. Winter concluded that the larger knee and hip moment variability during walking is attributable to the flexibility of the biarticular muscles (the "hamstrings" and the rectus femoris) that cross both joints. He also provided data showing the strong inverse relationship between joint moments at the hip and knee. As further evidence, he found that the CV for the sum of the knee and hip joint moments was very similar in value to CVs obtained for the ankle joint and support moments. A major result found in this study was that somewhat invariant kinematics were obtained for different walking trials when the knee and hip joint moments vary quite a bit. A limitation of this study, however, is that joint power values were not obtained.

One factor for which little information is available is how the kinetics are affected as the step length is varied while walking speed is held constant. Martin and Marsh (1992) performed such a study, manipulating the step length by 5% and 10% of leg length above and below the preferred step length. This had little effect on the vertical peaks of the GRF but changed the anterior-posterior (AP) peak force values by approximately 13% when step length was varied by 10%. While joint moments and power values were not obtained, one can get an idea of how much hip moments may have been affected by combining these results with the sensitivity analysis that Winter (1984) conducted. He reported that when the AP forces were varied by 10%, the average hip joint moments varied by 40% during walking. Winter's sensitivity analysis was for the case when AP force was varied but the kinematics were held constant, while in the Martin and Marsh study, the kinematics obviously varied as step length was altered and the speed was kept constant.

4.4 Accuracy

As previously indicated, several variables are involved in determining the kinetic measures during walking and other movements. These include the segment's inertial properties, the joint kinematics, the external forces (primarily the GRF) acting on the body, and the location on the body where these forces act. These variables often depend on other variables that may be directly measured or estimated and have a certain degree of accuracy. As an example, several anthropometric measures are often used to obtain estimates of the segment's inertial properties. Partly because of the variation in body structure characteristics among individuals, the inertial properties have traditionally been considered by biomechanics researchers to be accurate to within approximately 10% of their actual values. Other errors inherent in the data collection and processing steps would also exist. Examples of these include a determination of the joint center (or most appropriate joint center when the true joint center position varies), the limited resolution of data collection equipment (e.g., motion analysis equipment may only be accurate to approximately 1 cm), and the smoothing of the kinematic data. There have been very few studies (McCaw & DeVita, 1995; Challis & Kerwin, 1996; Nagano et al., 2000), however, in which specific variables were manipulated to determine the error they imparted to kinetic output variables.

Considering the generally nonlinear dependence of the joint moment and power values on several other variables, it would be difficult to determine the true accuracy of the kinetic measures. The accuracy of the input variables is generally between 0% and 20% of their actual values. Of course, inaccuracies in the data collection and processing steps can produce compensating errors that would reduce the overall inaccuracy of the kinetic output variables. However, it is reasonable to conclude that the accuracy of the desired joint moment and power measures would also be within this range. Therefore, one should keep in mind that the overall accuracy of many biomechanical variables and the kinetic results presented in this report may be as much as 20% different from their true values. The variability in the kinetic data along with the accuracy with which they can be determined provide the exoskeleton hardware designers with some measure of the tolerance in determining power requirements for various movements. Thus, the human power requirements indicated later in this report, combined with the variability and accuracy values stated previously, provide overall ranges of power for these respective movements.

5. Level of Effort Required During Locomotion

An important issue to consider during locomotion has to do with the level of effort that an individual must exert, that is, what are the required forces relative to the maximum forces that the individual is capable of developing in the various muscles used for locomotion? These relative force levels have strong implications regarding the likelihood of fatigue and the ability to meet particular locomotion requirements such as moving at high speed or traversing a steep grade. In essence, it would be desirable to know the maximum strength of individual muscle groups and the relative proportion of that strength that a subject is required to elicit during locomotion.

The standard method to measure strength of a particular muscle group is to use an isokinetic dynamometer, which has a rigid arm that is programmed to rotate about a fixed axis at a constant angular speed (between approximately 30 and 300 degrees per second) in a plane perpendicular to the axis of rotation. To test the strength of the muscle group, the subject's limb is rigidly attached to the dynamometer arm, and the center of the joint spanned by the muscle group is made to coincide with the arm's axis of rotation. The subject exerts maximal muscle effort while trying to rotate his or her joint in the same or opposite direction as the rotation of the dynamometer arm. While the subject is not able to alter the movement of the dynamometer arm, he or she is essentially exerting maximal concentric (shortening) or eccentric (lengthening) muscle action, respectively.

There are several caveats, however, in trying to compare moment and power measures obtained from isokinetic testing and joint moment and power values during locomotion. These can be

categorized in the areas of methodology, muscle mechanics, energy transfer, and neural factors. In terms of methodology, researchers usually do not smooth kinetic data, but it is probably appropriate to do so for movements in which subjects impact the ground, resulting in spikes in the GRF data. The impacts result in artificially high values for the calculated joint moments (near the instant of impact), which would not really exist in the muscle tendon complex because part of that force spike would be dissipated primarily by soft tissue within the body. In the area of muscle mechanics, the force-velocity (Hill, 1938) and length-tension (Gordon, Huxley, & Julian, 1966) relationships and force enhancement (Edman, Elzinga, & Noble, 1978) would prevent objective comparisons of moment and power measures between locomotion and isokinetic studies. Specifically, the muscle velocities and lengths may be different when peak kinetic values occur for these two movement paradigms along with the muscles involved in locomotion undergoing previous active stretching.

Another factor that would not exist during the isokinetic testing but is present during multiarticular movements is the capability of power being transferred in a proximal to distal direction (Jacobs et al., 1996; van Soest, Schwab, Bobbert, & van Ingen Schenau, 1993). For movements such as locomotion and jumping, the distal muscles have an increased moment and power-generating capability during the latter phase of ground contact. This is because monoarticular muscles acting about the more proximal joint provide power to an antagonist biarticular muscle that is simultaneously extending a more distal joint. A final factor that could potentially reduce the maximum muscle forces during isokinetic dynamometer testing is the possibility that neural inhibition may be present during isokinetic testing, resulting in a less than maximal muscle effort being provided by the neuromuscular control system. Combining all these factors provides ample reason to explain why locomotion is capable of producing larger joint moment and power values than those obtained in isokinetic dynamometer studies. Thus, the maximum moment or power results from isokinetic dynamometer studies should not be used to determine the degree of muscle activity during locomotion.

Table 2 provides data for a qualitative comparison of locomotion versus isokinetic moment values. Ranges of moment values for walking and running, along with values at specific angular velocities for isokinetic testing, are given for various muscle groups. Values in brackets represent average moment values based on the range of data obtained, and joint angular velocities are given in parentheses. For walking and running, the peak knee extensor moments are generally greater than the isokinetic moments, but they occur during an energy absorption phase when the muscles are lengthening and are capable of developing larger forces than when they are shortening (Hill, 1938). The large plantarflexor moment during walking and running is probably attributable to the muscles being close to their optimal force-producing length, the gastrocnemius' shortening velocity is near zero, the existence of force enhancement, and the distal transfer of power. For the joint power values (see Table 3), similar conclusions can be stated when one compares locomotion and isokinetic data. The differences in kinetic measures from locomotion studies and isokinetic studies occur because of differences in the methodology,

muscle mechanics, energy transfer, and neural factors. Therefore, actuator and power supply design criteria should be based on data from locomotion studies, not isokinetic studies.

Table 2. Maximum joint moments during walking, running, and concentric isokinetic testing

Joint	Muscle Group	Walking	Running	Isokinetic
Hip	Extensors	15-140 [100]	40-80	300
	Flexors	40-120 [70]		170
Knee	Extensors	5-140 [80]	125-273	235 (50°/s), 166 (60°/s), 154 (180°/s), 120 (400°/s)
	Flexors	15-50 [30]		93 (60°/s), 70 (180°/s)
Ankle	Plantarflexors	85-165 [130]	180-240	89 (30°/s), 50 (90°/s), 21 (180°/s)

Note: Moments given in newton meters; average peak values in brackets; angular velocities in parentheses. For walking, knee angular velocity is ≈ 100°/s, and ankle angular velocity is ≈ 50°/s at the time when peak moment occurs.

Table 3. Maximum joint powers generated during walking, running, and concentric isokinetic testing

Joint	Muscle Group	Walking	Running	Isokinetic
Hip	Extensors	0-175	160-660	200 (100°/s)
Knee	Extensors	10-235	210-1050	205 (50°/s), 840 (400°/s)
	Flexors	10-50		97 (60°/s), 220 (180°/s)
Ankle	Plantarflexors	180-790	550-1580	47 (30°/s), 79 (90°/s), 66 (180°/s)

Note: Powers given in watts; angular velocities in parentheses. For walking, ankle angular velocity is ≈ 200°/s at the time when peak power occurs.

6. Lower Limb Kinematics and Kinetics for Various Activities

6.1 Walking

6.1.1 Walking at a Normal Speed

The biomechanics of "normal" walking at a natural pace are well established. Kinematic (Kadaba, Ramakrishnan, & Wootten, 1990) and normalized (to body mass) kinetic (Eng & Winter, 1995) data for all three planes of motion are given in Appendix A (Figures A-1, A-2, and A-3 for the hip, knee, and ankle, respectively). Examining the joint angle data, one finds that the

greatest range of motion in the sagittal plane occurs at the knee (≈ 60°), followed by the hip (≈ 40°), and finally the ankle (≈ 25°). In the frontal plane, however, the hip exhibits the greatest range of motion (≈ 10°), while motion of the knee is around 6° and that of the ankle is nearly zero. The range of motion for all three joints in the transverse plane is approximately 10°.

With respect to sagittal plane joint moments, the largest peak extensor moment occurs at the ankle before toe-off (≈ 47% gait cycle), meanwhile, the largest peak flexor moment occurs almost simultaneously at the hip (≈ 50% gait cycle). During this phase of the locomotion cycle, the foot is pushing off from the ground and the ankle plantarflexor muscles are providing a forward thrust. Frontal plane joint moments at the ankle, as well as transverse plane joint moments for all three joints, appear to be very small in comparison with those of the sagittal plane. This is not the case, however, for the nearly equal peak knee and hip abductor moments that occur just after contralateral toe-off (≈ 10% gait cycle). While the peak hip abductor moment is comparable to the peak hip extensor moment, the peak knee abductor moment is more than twice as large as the peak knee extensor moment, indicating the necessity of conducting three-dimensional analyses of joint kinetics instead of the typical sagittal plane analysis. In addition, such locomotion biomechanics results are particularly useful in that they are quite helpful in determining the necessary design requirements for the exoskeleton.

When the power data are examined, some different trends exist as compared to the moment data when one considers kinetic measures about the three orthogonal axes during walking. Specifically, the angular velocities are larger in the sagittal plane versus the other two planes, resulting in proportionately larger power values for this primary plane of movement. For example, knee and hip peak power generated in the frontal plane is only one-third the respective sagittal plane values, while the knee and hip peak moments in the frontal plane are larger than their respective values in the sagittal plane. Thus, for normal speed walking, it is more important to conduct a three-dimensional analysis when one is considering joint moments, as compared to joint powers. In terms of describing the joint power results in Appendix A, in the sagittal plane, the largest peak power generated is at the ankle (just before toe-off), while the largest peak power absorbed is at the knee. Both of these occur in the sagittal plane. The peak power absorbed at the hip in the frontal and sagittal planes is approximately equal. Similar to the joint moment results, the joint powers for the ankle in the frontal plane and for all three joints in the transverse plane were quite small. During normal walking, toe-off occurs at approximately 60% of the gait cycle, and it is evident from these figures that the moments and powers are generally small during the swing phase (60% to 100% gait cycle). An exception to this is the power absorbed at the knee (see Figure A-2) in the sagittal plane since the hamstring muscles provide a braking action to knee extension before heel contact. A summary of the peak kinetic values associated with all three planes of motion for each joint is given in Table 4.

Table 4. Peak kinetic values during one cycle of normal walking

Variable	Plane	Action	Hip	Knee	Ankle
Moment (Nm/kg)	Sagittal	Extensor	1.15 (0.30)	0.46 (0.35)	1.73 (0.22)
		Flexor	1.10 (0.30)	0.43 (0.18)	0.20 (0.10)
	Frontal	Abductor	1.20 (0.25)	1.10 (0.20)	0.13 (0.10)
		Adductor	0.10 (0.10)	*	0.04 (0.04)
	Transverse	Inversion	0.20 (0.04)	0.11 (0.03)	*
		Eversion	0.20 (0.06)	0.09 (0.05)	0.10 (0.04)
Power (W/kg)	Sagittal	Generation	1.80 (0.50)	0.60 (0.50)	4.40 (1.10)
		Absorption	1.0 (0.55)	1.50 (0.50)	0.50 (0.30)
	Frontal	Generation	0.55 (0.15)	0.18 (0.13)	0.07 (0.06)
		Absorption	0.90 (0.60)	0.18 (0.12)	0.12 (0.08)
	Transverse	Generation	0.02 (0.06)	0.04 (0.02)	0.01 (0.03)
		Absorption	0.17 (0.16)	0.15 (0.10)	0.02 (0.02)

Note: Values in parentheses are ± one standard deviation.
*Indicates value ≈ 0.

6.1.2 Walking at Various Speeds

The effects of walking speed on lower extremity biomechanics have been reported for a study conducted by Winter (1983). Sagittal plane joint angles and powers were measured and calculated for all three joints at slow (85 steps/min), natural (105 steps/min), and fast (122 steps/min) cadences. Plots of these data are presented in Figure A-4. The point at which toe-off occurs in the gait cycle was found to be unaffected by change in walking speed, occurring at approximately 63% of the gait cycle for all three cadences. Although the joint angles for all three joints are seen to vary slightly with change in walking speed, these differences are considered to be very small when compared to the range of motion of each joint. With respect to joint power, the timing of transitions from power generation to absorption or absorption to generation was generally unaffected by change in walking speed; however, a high correlation between peak joint

powers and walking velocity was observed for all three joints. Several power bursts have been identified at the hip, knee, and ankle joints during one cycle of normal walking (see Figure A-4). Peak values occurring during these power bursts are presented in Table 5. An increase in peak joint power was shown to occur with increased walking speed for all power bursts except H1.

Table 5. Peak power values during one cycle of walking at three different cadences

Cadence		Hip			Knee			Ankle	
(steps/min)	H1*	H2	H3	K1	K2	K3	K4	A1	A2
85	0.16	-0.15	0.33	-0.35	0.10	-0.60	-0.51	-0.48	2.08
105	0.31	-0.25	0.68	-0.60	0.35	-0.70	-0.85	-0.50	3.33
122	0.26	-0.90	1.39	-2.10	1.08	-1.65	-1.30	-0.60	5.00

Note: Powers given in watts per kilogram body mass. Positive values indicate power generated; negative values indicate power absorbed.
*The locations of each power burst within the gait cycle are indicated in Figure A-4.

6.1.3 Walking With Loads

In 2000, Harman, Hoon, Frykman, and Pandorf reported about the effects of load carriage on lower extremity biomechanics during walking. Joint angle data were collected and joint moments were calculated for carried backpack loads of 6, 20, 33, and 47 kg while subjects walked at a speed of approximately 1.33 m/s; plots of these data are given in Figure A-5. In contrast to change in walking speed, the instant when toe-off occurs in the gait cycle was affected by change in carried load. As carried load increased from 6 to 47 kg, the duration of the stance phase was observed to increase from approximately 63.4% to 65.2% of the gait cycle. Timing of transitions from flexion to extension and extension to flexion also appears to be affected by change in carried load. The effect of change in carried load on hip joint angles was not reported, but slight changes in knee and ankle joint angles were. Peak knee flexion during mid-stance (≈ 10% to 30% gait cycle) was found to increase from approximately 22.5 to 27.5 degrees, while peak knee flexion at the transition from initial to mid-swing (≈ 72% gait cycle) was found to decrease from approximately 68 to 64 degrees with an increase in carried load. At the ankle, peak dorsiflexion during terminal stance (≈ 30% to 50% gait cycle) was found to decrease from approximately 11.5 to 10 degrees, and peak plantarflexion at the transition from mid- to terminal swing (≈ 90% gait cycle) was found to decrease from approximately 5 to 3.5 degrees with an increase in carried load. As with the joint angles, timing of transitions from extensor to flexor and flexor to extensor moments, as well as peak values obtained at each joint, appears to be affected by change in carried load. At the hip, peak extensor moment values during loading response, as well as peak flexor moment values during terminal stance, were found to increase with an increase in carried

load. Peak knee extensor moment values during mid-stance and peak ankle plantarflexor moment values during terminal stance were also found to increase with an increase in carried load, while peak knee flexor and ankle dorsiflexor moments did not follow a monotonically increasing trend. The peak extensor and flexor moment values obtained at each joint under each of the four different backpack loads are summarized in Table 6.

Table 6. Peak moment values during one cycle of walking with four different backpack loads

Load	Hip		Knee		Ankle	
(kg)	Extensor	Flexor	Extensor	Flexor	Plantarflexor	Dorsiflexor
6	0.81	0.78	0.72	0.34	1.76	0.13
20	0.85	0.88	0.81	0.40	2.08	0.10
33	0.88	1.12	1.20	0.33	2.15	0.20
47	1.00	1.24	1.37	0.35	2.41	0.16

Note: Moments given in newton meters per kilogram body mass.

As previously mentioned, the mass, size, and inertial properties of the exoskeleton are assumed to be equivalent to those of a human. Because humans vary in these dimensions, exoskeletons must be designed to fit a range of soldiers. Typically, equipment is developed to fit soldiers from the 5th to 95th percentile for a particular body dimension. For the analyses in this report, body mass is the dimension that will be used because of the assumption that joint moments and joint powers scale linearly with mass. Also, the body masses of male soldiers will be used because according to the DARPA BAA, the exoskeleton is initially being developed for combat soldiers.

According to the anthropometric survey of U.S. Army personnel conducted in 1988 (Gordon et al., 1989), the body masses of 5th and 95th percentile male soldiers are 61.59 and 98.07 kg, respectively. For normal walking, we obtain estimates of the range of peak sagittal plane kinetic values for male soldiers from the 5th to 95th percentile (by mass) by multiplying the given body mass by the moment and power values given in Table 4. The resulting ranges of peak moment and power values obtained by these calculations are presented in Table 7.

Table 7. Range of peak kinetic values for 5th to 95th percentile male soldiers during one cycle of normal walking

Variable	Action	Hip	Knee	Ankle
Moment				
(Nm)	Extensor	71 to 113	28 to 45	106 to 169
	Flexor	68 to 108	26 to 42	12 to 20
Power				
(W)	Generation	111 to 177	37 to 59	271 to 432
	Absorption	62 to 98	92 to 147	31 to 49

6.2 Running

Sagittal plane kinetic data for the stance phase of "normal" running at a moderate pace of approximately 3.8 m/s have been published by DeVita, Torry, Glover, and Speroni (1996); plots of these data are shown in Figure A-6. Peak moment and power values associated with these data are summarized in Table 8. When the joint moment data are examined, it can be seen that, similar to walking, the largest peak extensor moment occurs at the ankle (in this case, near mid-stance, ≈ 50% stance phase), while the largest peak flexor moment again occurs at the hip. With respect to joint power, the largest peak power generation again occurs at the ankle, and the largest peak power absorption value is found at the knee during the loading response (≈ 10% to 30% stance phase). With the exceptions of peak hip and ankle flexor moments, all the peak kinetic values for running at a moderate pace are much higher than those for walking at a natural pace (see Table 4).

Table 8. Peak kinetic values during stance phase of normal running at 3.8 m/s

Variable	Action	Hip	Knee	Ankle
Moment				
(Nm/kg)	Extensor	2.81 (0.30)	2.74 (0.38)	3.34 (0.12)
	Flexor	0.61 (0.23)	0.53 (0.00)	*
Power				
(W/kg)	Generation	3.80 (1.52)	12.9 (2.08)	17.5 (0.76)
	Absorption	11.0 (4.18)	18.2 (1.52)	12.2 (0.69)

Note: Values in parentheses are ± one standard deviation.
* indicates a value ≈ 0.

Arampatzis, Brüggemann, and Metzler (1999) have reported about the effects of change in speed on lower limb biomechanics during the stance phase of running. Sagittal plane data were collected at the knee and ankle for subjects running at five speeds ranging from jogging (2.61 m/s) to sprinting (6.59 m/s), and the resulting joint moments and powers occurring during the stance phase were calculated. Plots of these results are shown in Figure A-7, and a summary of the peak moment and power values obtained is provided in Table 9. Similar to the effects of change in speed on walking biomechanics, all the peak kinetic values during the stance phase of running are shown to increase with increased speed.

As with walking, an estimate of the peak kinetic values for the range of 5th to 95th percentile male soldiers running at a moderate pace can be obtained. This is done by multiplying the corresponding body masses with the mean normalized joint moment and power data for running, as reported in Table 8. The resulting values are presented in Table 10.

Table 9. Peak kinetic values during the stance phase of running at five different speeds

	Knee			Ankle		
	Moment	Power		Moment	Power	
Speed (m/s)	Extensor	Generated	Absorbed	Extensor	Generated	Absorbed
2.61	2.00	4.23	7.75	2.45	6.60	3.30
3.55	2.57	5.62	12.0	2.78	10.1	4.60
4.47	2.64	6.92	13.2	2.99	12.3	5.90
5.60	2.74	9.89	14.2	3.20	16.4	8.70
6.59	2.99	10.7	16.8	3.44	21.0	12.2

Note: Moments given in newton meters per kilogram; powers given in watts per kilogram.

Table 10. Peak moment values for 5th to 95th percentile male soldiers during one cycle of running at a moderate pace (3.8 m/s)

Variable	Action	Hip	Knee	Ankle
Moment (Nm)	Extensor	173 to 276	168 to 268	206 to 328
	Flexor	37 to 60	33 to 52	*
Power (W)	Generation	234 to 373	796 to 1267	1076 to 1714
	Absorption	679 to 1081	1123 to 1789	749 to 1192

Note: * indicates a value ≈ 0.

6.3 Stair Climbing

In 1980, Andriacchi, Anderson, Fermier, Stern, and Galante performed an analysis of stair ascent and descent, reporting the stance phase joint moments for all three planes of motion. In this study, subjects ascended and descended a three-step staircase, and kinematic data were collected for one stride between the first and third steps. Joint moments were calculated during the stance phase for all three planes of motion at each joint, and these resulting data are shown in Figure A-8. A summary of the peak joint moment values obtained for each plane of motion is presented in Table 11. During the stance phase of stair ascent, the largest joint moments appear to have occurred in the sagittal plane, but peak abductor moments in the frontal plane are also significant, with values equal to approximately one-half to three-quarters of their corresponding peak extensor moments. Similarly, during the stance phase of stair descent, the largest peak joint moments again appear to have occurred in the sagittal plane, while peak abductor moments range in value from approximately one-half to two-thirds of their corresponding peak extensor moment. Transverse moments during the stance phase of both stair ascent and descent appear to be rather small for all three joints.

Table 11. Three-dimensional peak moment values during the stance phase of normal stair ascent and descent

	Plane	Action	Hip	Knee	Ankle
Ascent	Sagittal	Extensor	1.27	0.89	1.34
		Flexor	*	0.28	*
	Frontal	Abductor	0.63	0.77	0.60
		Adductor	0.05	*	*
	Transverse	Inversion	0.03	0.08	0.02
		Eversion	0.21	0.05	0.10
Descent	Sagittal	Extensor	0.99	1.55	1.20
		Flexor	0.35	0.70	*
	Frontal	Abductor	0.65	0.44	0.63
		Adductor	0.14	0.14	*
	Transverse	Inversion	0.04	0.13	0.03
		Eversion	0.21	0.03	0.06

Note: Moments given in newton meters per kilogram
* indicates a value ≈ 0.

Sagittal plane lower extremity joint powers have been published by Duncan, Kowalk, and Vaughan (1997) for normal stair ascent and descent. In their study, subjects also ascended and descended a three-step staircase, kinematic data were collected for each lower extremity joint, and joint powers were calculated with a six-degree-of-freedom approach. Plots of the resulting joint powers are shown in Figure A-9, and a summary of the peak power values obtained for

each joint is given in Table 12. During both stair ascent and descent, the largest peak power generation appears to have occurred at the ankle, while the largest peak power absorption during stair ascent occurred at the knee and during stair descent at the ankle. In contrast to walking, during stair ascent, peak power generation at the knee is approximately four times greater, while peak power generation at the ankle decreases by about one-third. Peak power absorption values at all three joints are also reduced by one-half to three-fourths in comparison to those values seen during walking. The most notable difference between normal walking and stair ascent is the role reversal of the knee joint from power absorber to power generator. During stair descent, peak power generation values at all three joints and peak power absorption at the hip are also reduced to approximately one-third to one-half of their corresponding values for normal walking, while peak power absorption values at the knee and ankle are approximately one and one-third and five times larger, respectively. The role of the ankle joint is split nearly evenly between power generation and absorption, rather than acting solely as a power generator as seen in normal walking.

Table 12. Peak power values during one cycle of normal stair ascent and descent

	Action	Hip	Knee	Ankle
Ascent	Generation	1.50	2.10	2.90
	Absorption	0.50	0.70	0.10
Descent	Generation	0.65	0.25	2.05
	Absorption	0.50	2.00	2.40

Note: Powers given in watts per kilogram.

Sagittal plane peak kinetic values during stair ascent and descent for the range of 5th to 95th percentile male soldiers can again be estimated through multiplication of the appropriate body masses with the normalized sagittal plane joint moment data reported in Table 11 and the normalized joint power data reported in Table 12. The resulting values are presented in Table 13.

Table 13. Peak kinetic values for 5th to 95th percentile male soldiers during one cycle of stair ascent and descent

	Variable	Action	Hip	Knee	Ankle
Ascent	Moment	Extensor	78 to 125	55 to 87	83 to 131
	(Nm)	Flexor	*	17 to 27	*
	Power	Generation	92 to 147	129 to 206	179 to 284
	(W)	Absorption	31 to 49	43 to 69	6 to 10
Descent	Moment	Extensor	61 to 97	95 to 152	74 to 118
	(Nm)	Flexor	22 to 34	43 to 69	*
	Power	Generation	40 to 64	15 to 25	123 to 195
	(W)	Absorption	31 to 49	123 to 196	148 to 235

Note: * indicates a value ≈ 0.

6.4 Jumping

The lower extremity kinetics of a running vertical jump with a single-legged take-off have been reported by Stefanyshyn and Nigg (1998). In this study, five subjects performed running vertical jumps, and kinematic data were collected for the stance phase of the take-off leg before toe-off. Plots of the averaged data from this report are shown in Figure A-10, where the normalized stance phase represents time periods ranging from 0.23 to 0.35 seconds, 0% stance phase is the instant of foot contact with the floor, and 100% stance phase is the instant of toe-off. The peak kinetic values obtained for each joint during the running vertical jump are summarized in Table 14. During the running vertical jump, it appears that the largest peak extensor moment occurs at the hip, while the largest peak flexor moment occurs at the knee. With respect to joint power, the largest peak generation during the running vertical jump occurs at the ankle, and the largest peak power absorption appears to occur at the ankle.

Table 14. Peak kinetic values during the stance phase before toe-off of a running vertical jump

Variable	Action	Hip	Knee	Ankle
Moment	Extensor	4.18 (1.15)	2.77 (0.72)	4.08 (0.59)
(Nm/kg)	Flexor	0.35 (1.47)	1.07 (0.45)	0.09 (0.29)
Power	Generation	9.24 (6.48)	10.7 (8.15)	26.8 (3.33)
(W/kg)	Absorption	3.02 (0.56)	6.09 (5.31)	7.81 (3.74)

Note: Values in parentheses are ± one standard deviation.

We estimated the range of peak kinetic values for 5th to 95th percentile male soldiers during the take-off stance phase of a running vertical jump by multiplying their respective body masses by the values given in Table 14. The values resulting from this process are presented in Table 15.

Table 15. Peak kinetic values for 5th to 95th percentile male soldiers during the stance phase before toe-off of a running vertical jump

Variable	Action	Hip	Knee	Ankle
Moment (Nm)	Extensor	257 to 410	171 to 272	251 to 400
	Flexor	22 to 35	66 to 106	5 to 9
Power (W)	Generation	569 to 906	658 to 1048	1648 to 2624
	Absorption	186 to 297	375 to 598	481 to 766

6.5 Kneeling

Very little published biomechanics data exist for kneeling, but some information concerning range of motion and joint moments is presented in an American Society of Biomechanics (ASB) conference abstract (Nagura et al., 2000) and American Society of Mechanical Engineers (ASME) conference abstract (Nagura, Dyrby, Alexander, and Andriacchi, 2001). The ASB abstract discusses results for all three joints from a study in which subjects performed four tasks, single- and double-legged rises from a kneeling position, and single- and double-legged kneeling from a standing position. Ranges of motion for each joint (approximately 65.9 degrees at the hip, 149.7 degrees at the knee, and 88.6 degrees at the ankle) were similar across the four different tasks. These values are significantly higher than those seen during normal walking or stair climbing. During the single-legged rise, the largest peak extensor moment occurred at the hip ($\approx$ 1.4 Nm/kg), followed closely by the knee ($\approx$ 1.3 Nm/kg), and ankle ($\approx$ 0.8 Nm/kg). Peak flexor moments were negligible for all three joints during this task. During the double-legged rise, the largest peak extensor moment occurred at the knee ($\approx$ 2.4 Nm/kg), followed by the ankle ($\approx$ 1.4 Nm/kg), and hip ($\approx$ 0.2 Nm/kg). Peak flexor moments were again negligible at the knee and ankle, but the peak hip flexor moment was approximately 1.4 Nm/kg. Joint moments for the single- and double-legged kneeling tasks were reported to be similar to those for the single- and double-legged rising tasks. In the ASME abstract, range of motion and joint moments at the knee during the same four tasks are discussed. Range of motion at the knee is reported to vary from approximately 126 to 152 degrees, depending on the task. Peak knee extensor moments for the single-legged rising and kneeling tasks are reported to be approximately 1.0 and 1.1 Nm/kg, respectively, while peak knee extensor moments for the double-legged rising and kneeling tasks are reported to be approximately 2.3 and 2.1 Nm/kg, respectively.

7. Lower Limb Peak Power Profiles for Soldier Missions

7.1 Procedure

With the hip, knee, and ankle joint power data reported in biomechanical literature and a series of simple steps, lower extremity peak power profiles were generated for two soldier missions consisting of various tasks. Several assumptions were made in this process. Each of these five assumptions should be considered as potential sources of error in the values resulting from this process.

1. Increasing carried load has the same effect on peak joint power as does increasing body mass,

2. The power values observed in the frontal plane are generally small and in the transverse plane generally very small in comparison to those in the sagittal plane; therefore, only the power generated and absorbed in the sagittal plane was considered,

3. Normal gait is symmetric,

4. Based on range of motion and peak extensor moments, peak power values for assuming and getting out of kneeling and prone positions, crawling, and climbing a ladder can be approximated with values of stair ascent, and

5. Power values of other lower extremity joints, such as the metatarsophalangeal joint, are small and therefore not included in the calculation of peak power.

The first step in this process involved the development of mission spreadsheets containing time, task, and load information for each of the missions examined (see Appendix B, Table B-1). Estimations of the peak and average powers required for each of the task and load combinations were then made according to the method described in the following paragraph. Finally, the peak power values obtained for each task and load combination were entered into the mission spreadsheets appropriately, and peak power profiles were generated for each mission scenario. This process could be applied to any mission scenario that anyone would want to create.

Several steps were taken in order to estimate the peak and average powers required for each task and load combination. Based on the first assumption listed previously, we estimated the total power required during one cycle of each task and load combination by first multiplying the appropriate normalized hip, knee, and ankle joint power data for each task by the correct mass value. The mass values used in this calculation are as follow: 50th percentile male soldier ($\approx$ 77 kg) plus fighting load ($\approx$ 24 kg) or sustainment load ($\approx$ 35 kg), resulting in values of 101 and 112 kg, respectively. The mass of the 50th percentile male soldier is used to keep the analyses simple and to provide initial guidance. Table 16 shows the individual masses of each component included in the fighting and sustainment loads. Then, based on the second and third assumptions listed previously, the absolute joint power data for each task and load combination

(except for the running vertical jump) were plotted simultaneously for the hip, knee, and ankle joints of the left and right lower limbs, offsetting the curves of each leg by 50% of the gait cycle (see Figure B-1a). For the running vertical jump, the absolute joint power data were simultaneously plotted for the hip, knee, and ankle joints of the take-off leg only. The actuators of the exoskeleton will need to develop power during the power generation and power absorption phases of these tasks; therefore, the absolute values of the resulting joint powers for each task and load combination were taken. Next, a total power curve was obtained for each task and load combination via the equation

$$TP(t) = \sum_{i}^{N} \left| P_i(t) \right| ,$$

in which t is percent time in the gait cycle (i.e., between 0% and 100%), TP is the total power, P is the joint power, i indicates the joint (i.e., left or right hip, knee, or ankle), and N is the number of joints (three for running vertical jump; six for all other tasks). Finally, the peak power value was obtained through identification of the maximum value along the total power curve (see Figure B-1b), and the average power value was calculated for each task and load combination. These values are summarized in Table 17.

Table 16. Loads carried

Load	Mass (kilograms)
Fighting Load	
Battle dress uniform	1.7
Boots	1.5
Personal armored system for ground troops (PASGT) helmet	1.4
PASGT vest	5.4
M16A2	3.7
Bayonet	0.6
Load-carrying equipment (LCE)	0.7
Ammunition pouches (2) with 180 rounds in six magazines	3.2
Hand grenades (2)	0.9
Canteens (2) with water	2.9
Protective mask and hood	1.3
Poncho	0.7
TOTAL	*24.0*
Existence Load	
Army lightweight individual carrying equipment (ALICE) pack	1.1
Additional clothing	0.5
Personal hygiene kit	1.2
Food	1.3
Sleeping bag	3.4
Sleeping mat	1.6
Field Jacket and Gloves	1.9
TOTAL	*11.0*
Sustainment Load	
Fighting load + existence load	35.0

Table 17. Peak and average power values for typical soldier tasks

Task	Total Mass (soldier plus carried load) Fighting (101 kg)	Sustainment (112 kg)
Slow Walk (85 steps/min)	283 (119)	314 (131)
Natural Walk (105 steps/min)	411 (180)	456 (200)
Fast Walk (122 steps/min)	693 (315)	769 (349)
Jog (2.7 m/s)	1966 (1041)	2180 (1155)
Run (3.8 m/s)	3147 (1693)	3489 (1877)
Sprint (7.6 m/s)	5547 (3522)	6151 (3906)
Running Vertical Jump	3559 (1325)	3947 (1369)
Ascend Stairs	501 (229)	556 (254)
Descend Stairs	478 (270)	530 (300)

Note: All peak(average) power values are given in watts. All values represent absolute power for all six lower extremity joints (left and right hip, knee, and ankle joints) during the given task and load combination, except for jumping, which assumes a running jump with single-legged take-off.

7.2 Mission Scenarios

7.2.1 Movement to Contact

In the movement-to-contact mission, a platoon is moving through a rural area. They are carrying their sustainment loads (35 kg). At the beginning of the mission, the platoon marches and takes rest breaks according to Field Manual (FM) 21-18 (Department of the Army, 1990). When the platoon makes contact with the enemy, they drop their existence loads and carry only their fighting loads (24 kg). The platoon members who will engage the enemy use cover and concealment to move according to FM 7-8 (Department of the Army, 1992) and FM 21-75 (Department of the Army, 1984). This movement includes a series of rushes, that is, running for 3 to 5 seconds, going prone, crawling, rolling left or right, getting up and repeating the sequence until they are within approximately 100 yards of the enemy. They engage the enemy from covered positions and then move in another series of rushes to close with and destroy the enemy. They consolidate and reorganize. Then they remove casualties. After resting briefly, they return to pick up their existence loads. They repeat the tasks from the first part of the mission and engage another group of enemy soldiers. After they go back to pick up their existence loads this time, they march and take rest breaks until they find a safe place to rest for the night. The entire mission takes 16 hours, and the resulting peak power profile is shown in Figure B-2 for a 50th percentile male soldier. With the assumptions stated earlier concerning the exoskeleton's design, its power requirements can be estimated from Figure B-2 and assumed to be within 20% of their actual values. Thus, a power supply that is capable of producing a nearly constant mean value of approximately 200

±40 W, and peak values of 500 ±100 to 5500 ±1100 W at frequencies ranging from approximately 2 to 4 Hz, depending on the task, would be required for this mission.

7.2.2 Clear Building

In this scenario, a platoon is going to clear a building during daylight. The soldiers are carrying their fighting loads (24 kg). The building is three stories high and has a basement. Each floor is approximately 17,850 square feet. There are six to ten enemy soldiers in the building. The platoon follows FM 90-10-1 (Department of the Army, 1993) and FM 7-8 (Department of the Army, 1992). The platoon divides into squads. One squad provides security outside the building. Another squad is held in reserve. Two squads enter and clear the building. The two squads that enter and clear the building approach from the back. They use a ladder and enter a second floor room through a window. Then they work their way to the stairs. Inside the building, one squad is responsible for securing the stairs, and the other squad is responsible for clearing the rooms. The squad that clears the rooms divides into two fire teams and goes up to the third floor. One fire team provides security in the hallway while the other team clears rooms. A fire team clears about four to six rooms and then switches to security while the other team clears rooms. They alternate roles in this fashion, clearing the third floor, the second floor, the first floor, and finally, the basement. The entire mission takes 2 hours and 12.5 minutes, and the resulting peak power profile is shown in Figure B-3 for a 50th percentile male soldier on one of the room-clearing fire teams. As with the movement-to-contact mission, the exoskeleton's power requirements during a clear-building mission can be estimated from Figure B-3 and assumed to be within 20% of their actual values. Although this power profile varies much more than that of the previous scenario, it appears that a power supply capable of producing a mean power value between 100 ±20 and 300 ±60 W, and peak values of 400 ±80 to 3500 ±700 W at 2 to 4 Hz, depending upon the task, would be required for such a mission.

8. Conclusions

The goals for this work were to determine the angles, torques, and powers at the lower limb joints of an exoskeleton and to estimate the requirements for a system to power the lower limbs of that exoskeleton. It was assumed that the angles, torques, and powers could be estimated from biomechanical data collected from humans. Based on biomechanical data gathered from various published sources, we have provided estimates of angle, torque, and power values at the hip, knee, and ankle joints during various activities. Estimated peak power values were also determined for two hypothetical missions to produce lower limb peak power profiles. Keeping in mind the limitations and variability of kinematic and kinetic data and the assumptions made in the application of such data to an exoskeleton, these estimated values (see Appendix A, Table 17, and Figures B-2 and B-3) could be used to design lower limb joints and actuators and to size the power supply for an exoskeleton.

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Appendix A: Lower Limb Kinematics and Kinetics for Various Activities

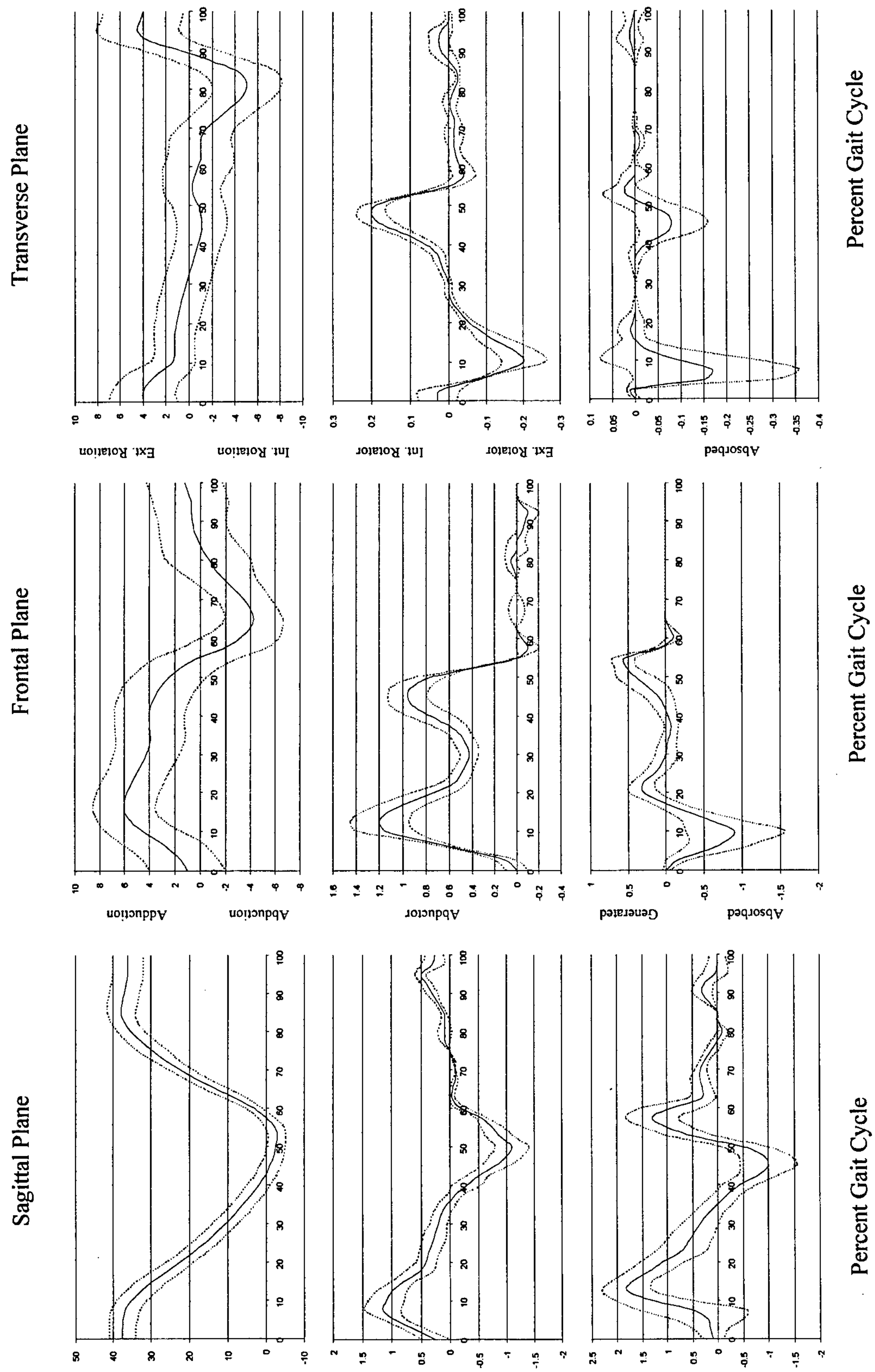

Figure A-1. Hip joint angles, moments, and powers in the sagittal, frontal, and transverse planes during one cycle of normal walking. (Solid lines are mean values and dotted lines are ± one standard deviation.)

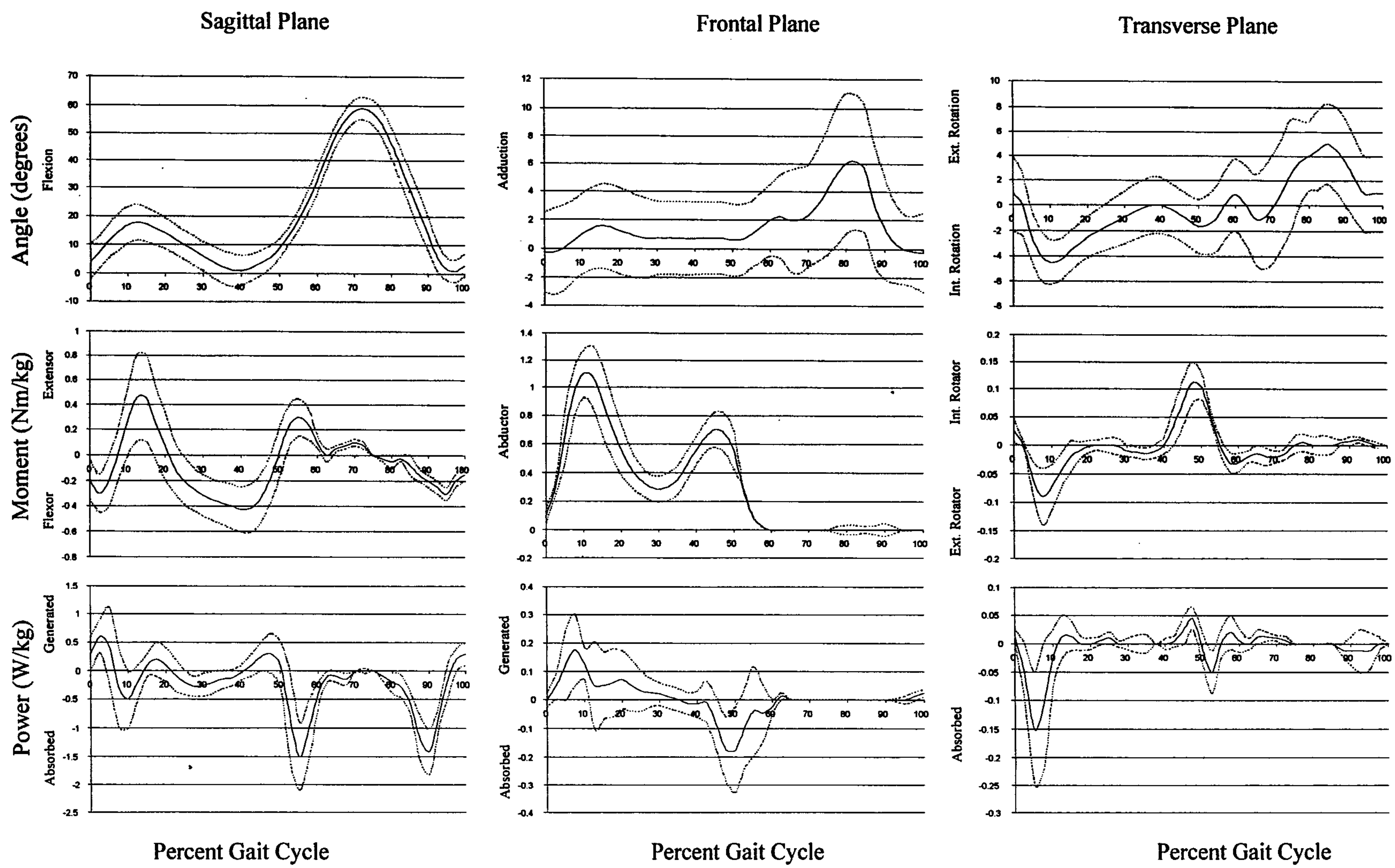

Figure A-2. Knee Joint angles, moments, and powers in the sagittal, frontal and transverse planes during one cycle of normal walking.

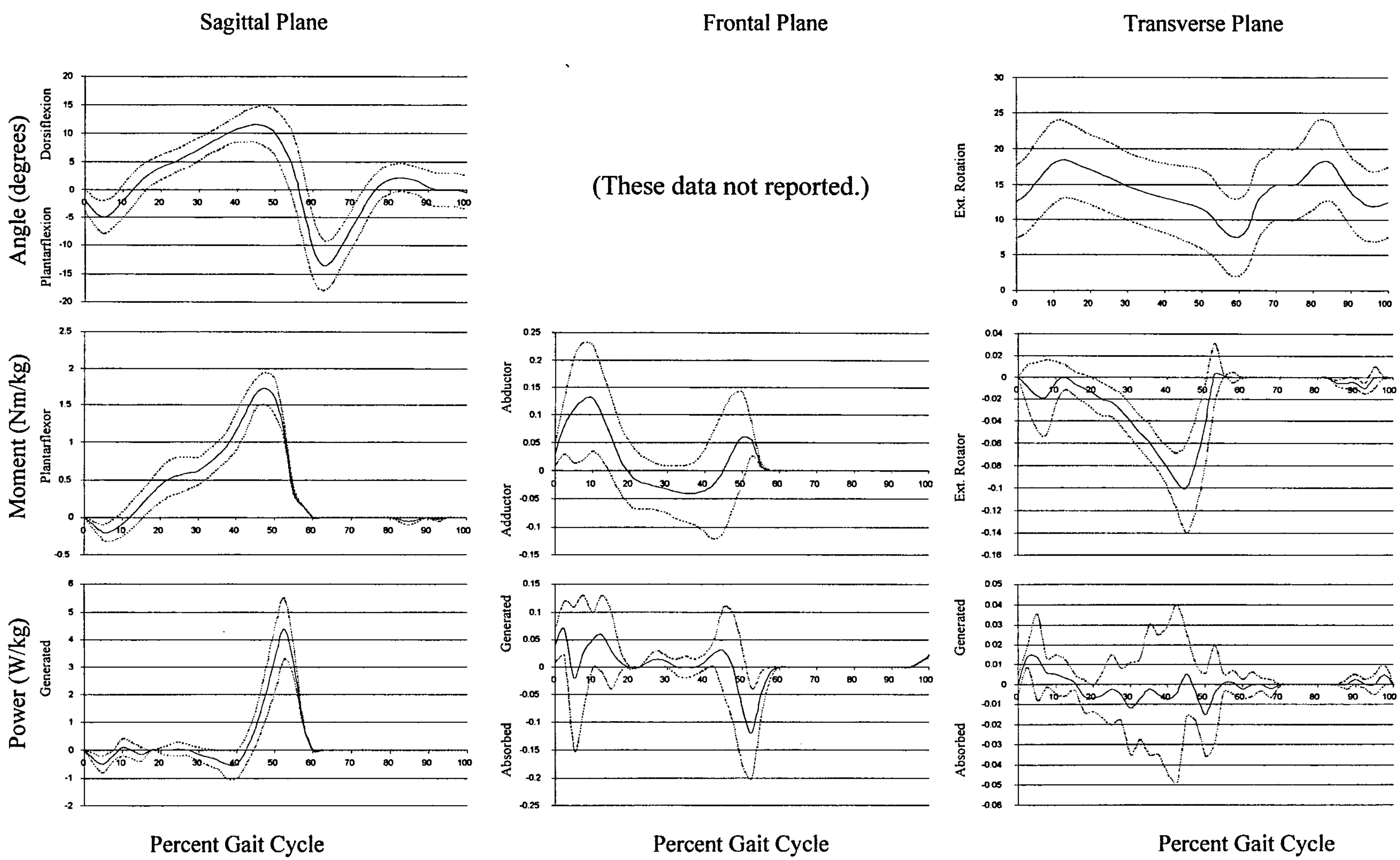

Figure A-3. Ankle joint angles, moments, and powers in the sagittal, frontal and transverse planes during one cycle of normal walking.

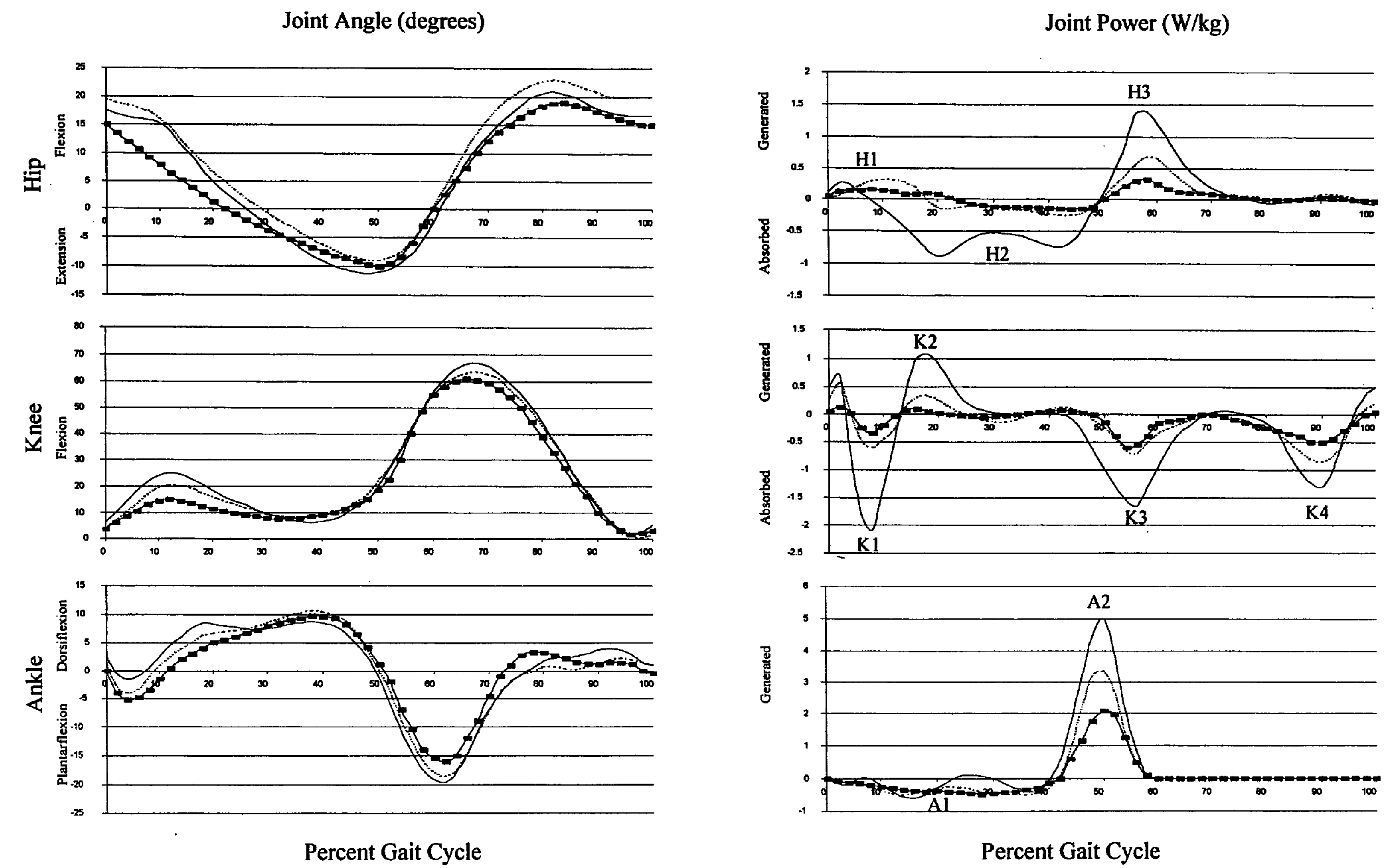

Figure A-4. Sagittal plane joint angles and powers during one cycle of walking at three cadences (—■— = slow, - - - = natural, —— = fast).

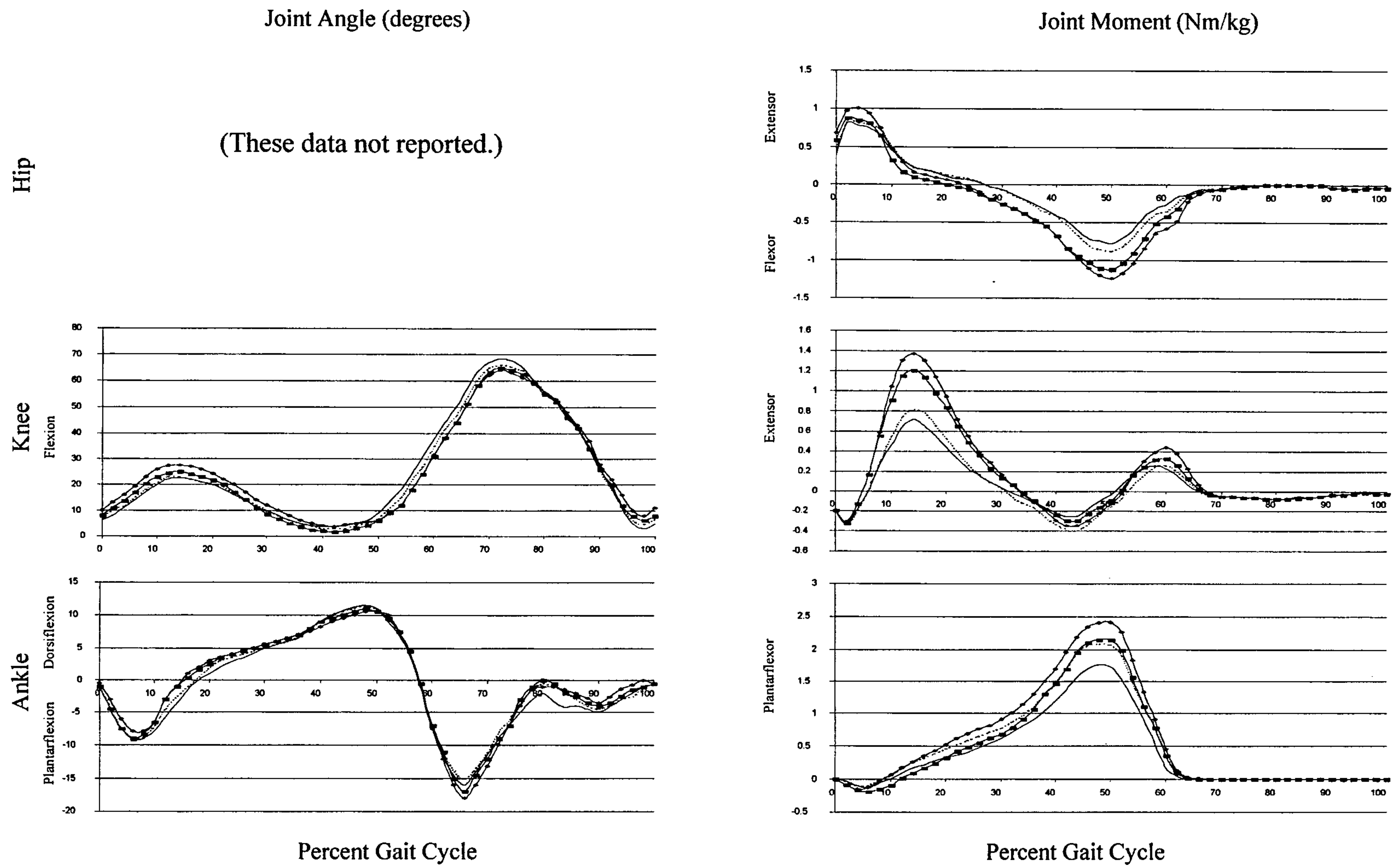

Figure A-5. Sagittal Plane Joint Angles and Moments During One Cycle of Walking at 1.33 m/s With Four Different Backpack Loads (—— = 6 kg,---- = 20 kg,—■— = 33 kg,—◆—= 47 kg).

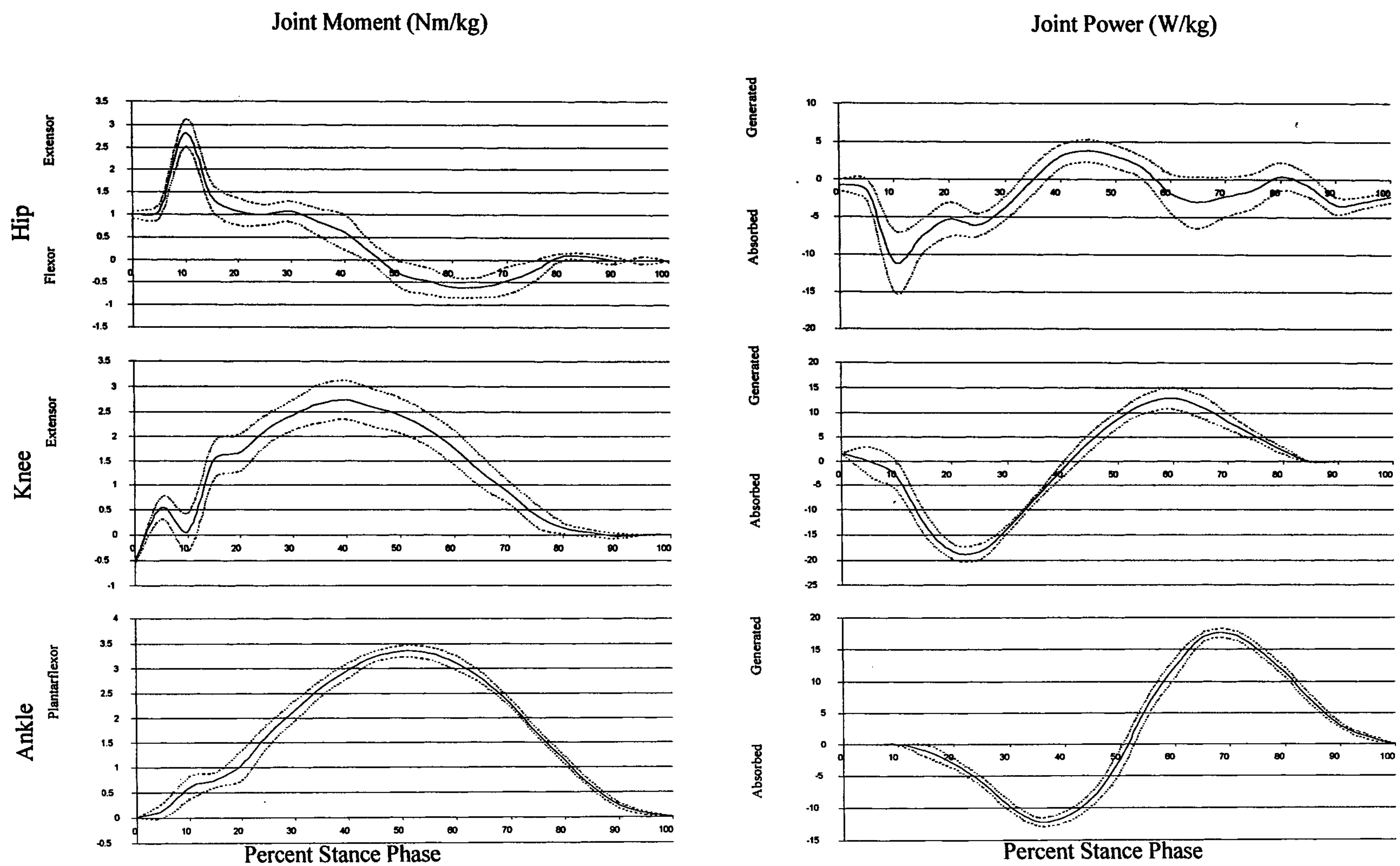

Figure A-6. Sagittal plane joint moments and powers during the stance phase of running at a moderate pace (3.8 m/s). (Solid lines are mean values and dotted lines are ± one standard deviation.)

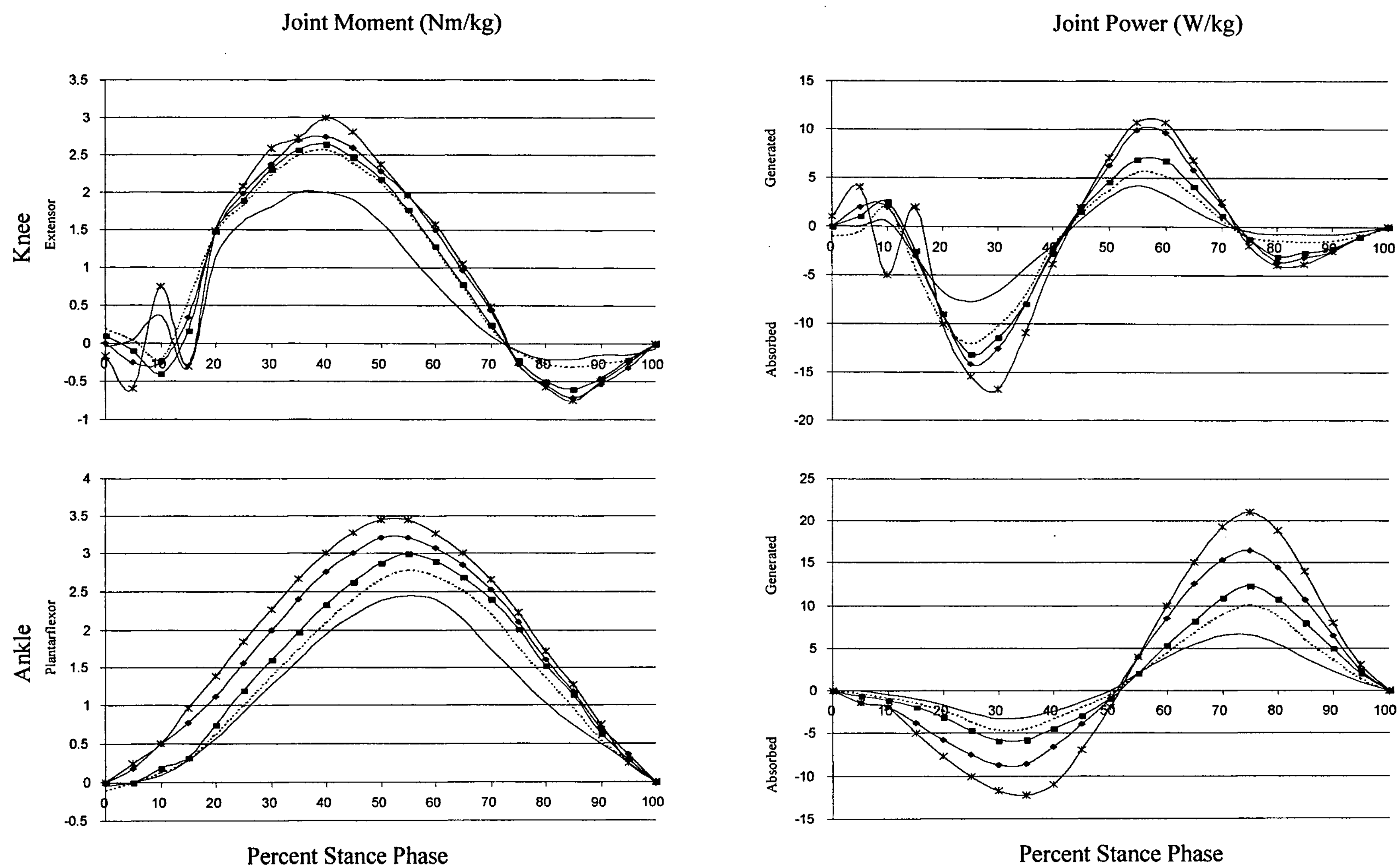

Figure A-7. Sagittal plane joint moments and powers during the stance phase of running at five velocities (—— = 2.61 m/s, ---= 3.55 m/s, —■— = 4.47 m/s, —◆— = 5.60 m/s, —✳— = 6.59 m/s).

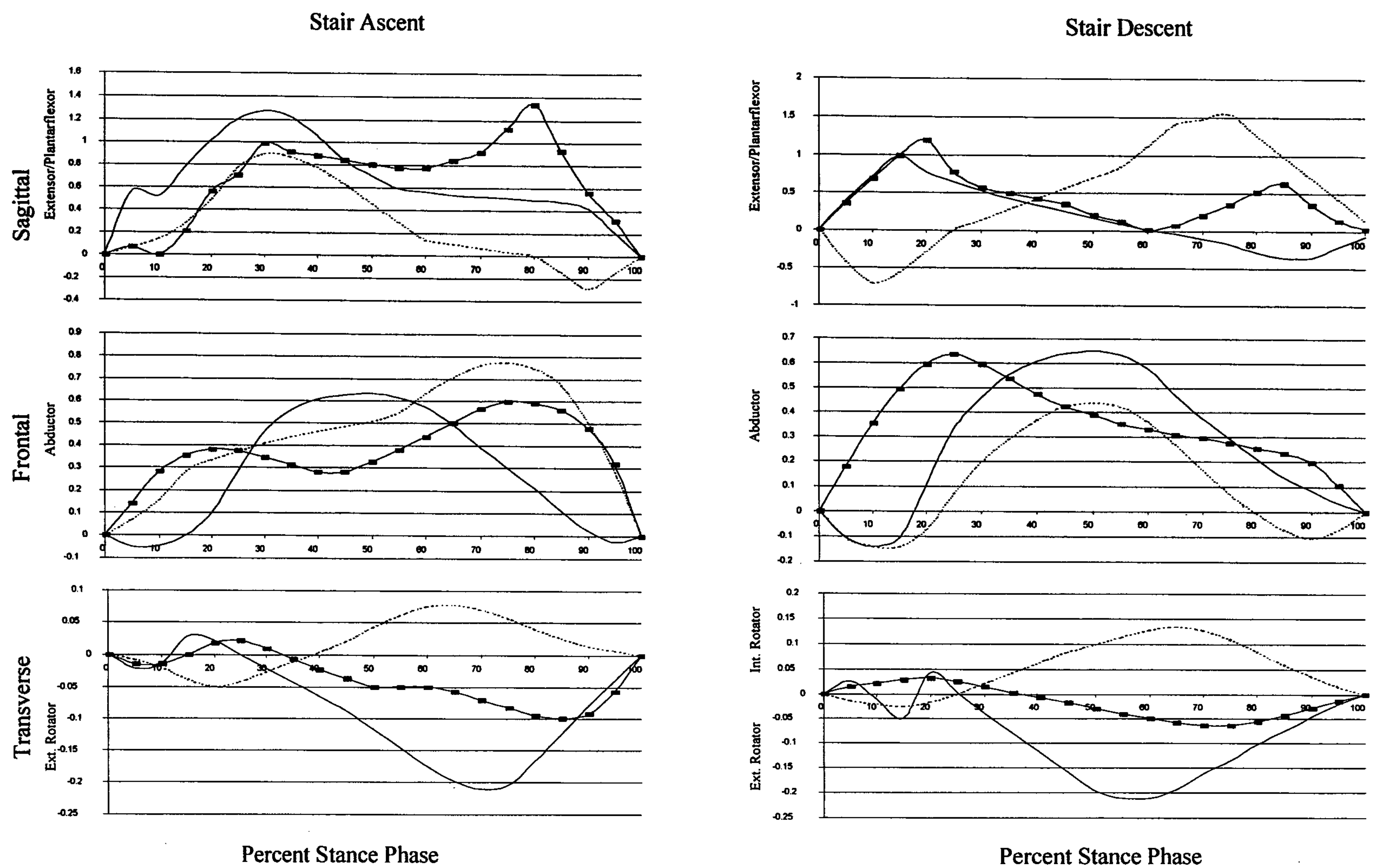

Figure A-8. Joint moments in the sagittal, frontal, and transverse planes during stance phase of stair ascent and descent.
(—— = hip,---- = knee,—■— = ankle). (Values are in newton meters per kilogram.)

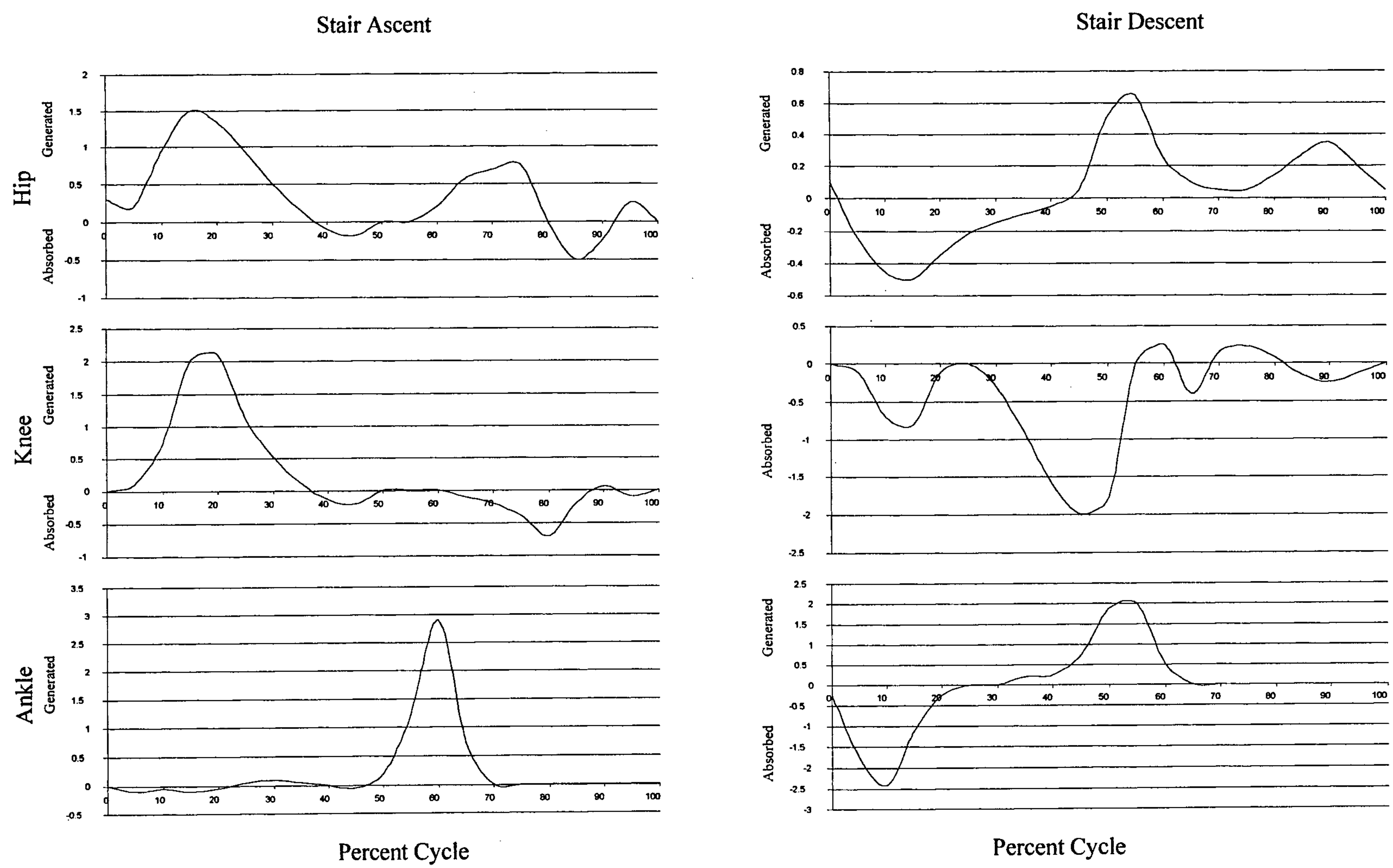

Figure A-9. Sagittal plane joint powers during one cycle of stair ascent and descent. (Values are in watts per kilogram.)

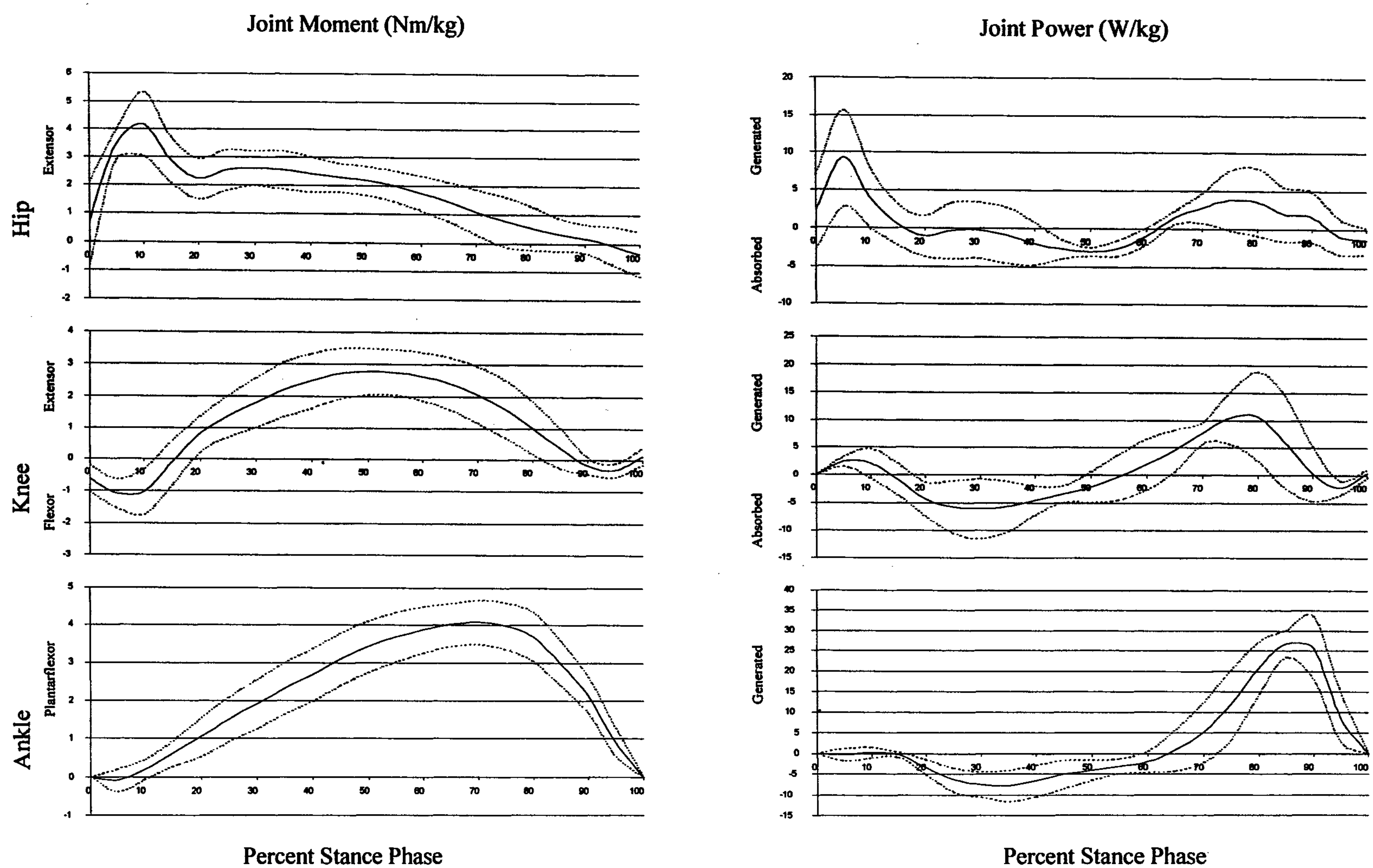

Figure A-10. Sagittal plane joint moments and power during the stance phase before toe-off of a running vertical jump. (Solid lines are mean values, and dotted lines are ± one standard deviation.)

Appendix B: Lower Limb Peak Power Profiles for Soldier Missions

Table B-1. Example of the peak total power spreadsheet set up for a typical soldier mission. (In this case, the soldier walks for 45 minutes with a sustainment load, rests for 15 minutes, walks for 30 minutes with a sustainment load, kneels and rests for 2 minutes, etc.)

Time (min)	Task	Load
0	Walking	Sustainment
45	Resting	
60	Walking	Sustainment
90	Kneeling and Resting	Sustainment
92	Walking	Sustainment
110	Resting	
120	Walking	Sustainment
130	Kneeling and Resting	Sustainment
134	Walking	Sustainment
150	Prone - Security	Sustainment
152	Walking	Sustainment
170	Resting	
180	Walking	Sustainment
200	Prone - Security	Sustainment
205	Walking	Sustainment
230	Resting	
240	Walking	Sustainment

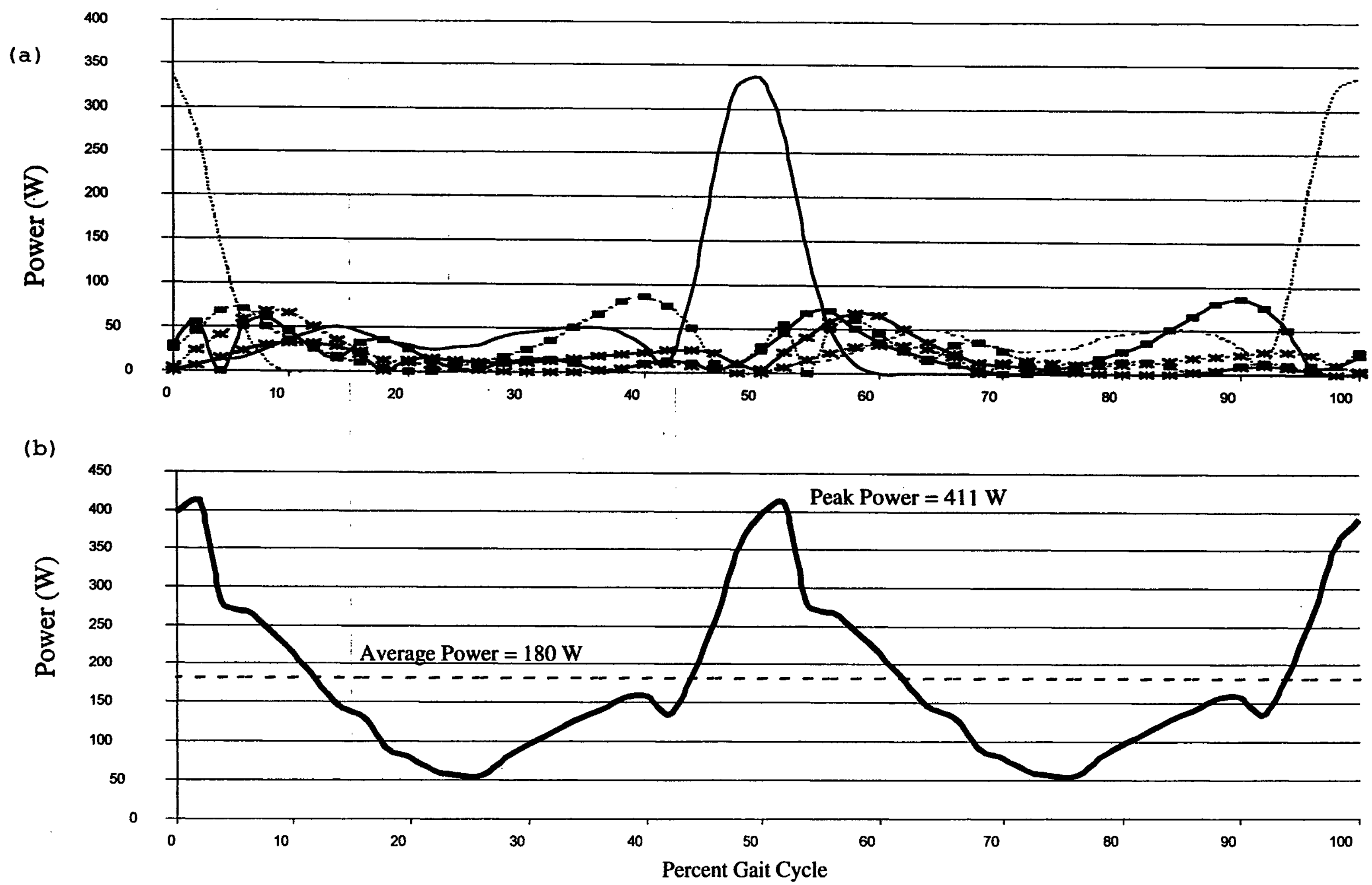

Figure B-1. Example of identification of peak and average power values during one cycle of a task and load combination (in this case, walking at a natural pace with a fighting load): (a) simultaneous plotting of absolute power curves for all six lower limb joints (—— = L. Ankle, --- = R. Ankle, —■— = L. Knee, -■- = R. Knee, —✱— = L. Hip, -✱- = R. Hip); (b) Total power curve (solid line) with peak and average power values identified.

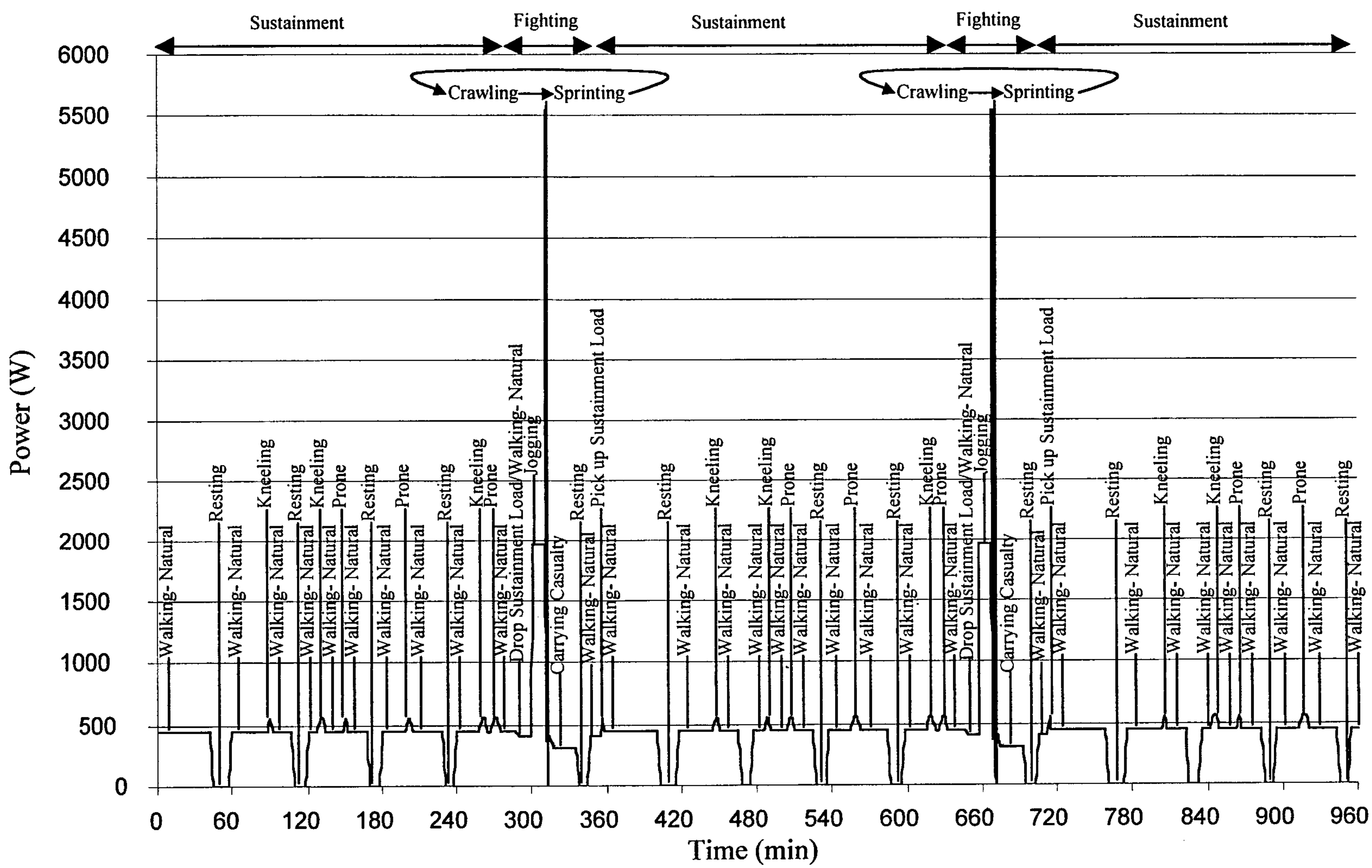

Figure B-2. Lower limb peak power profile for movement-to-contact mission scenario. (Carried load is indicated by text and arrows above plot.)

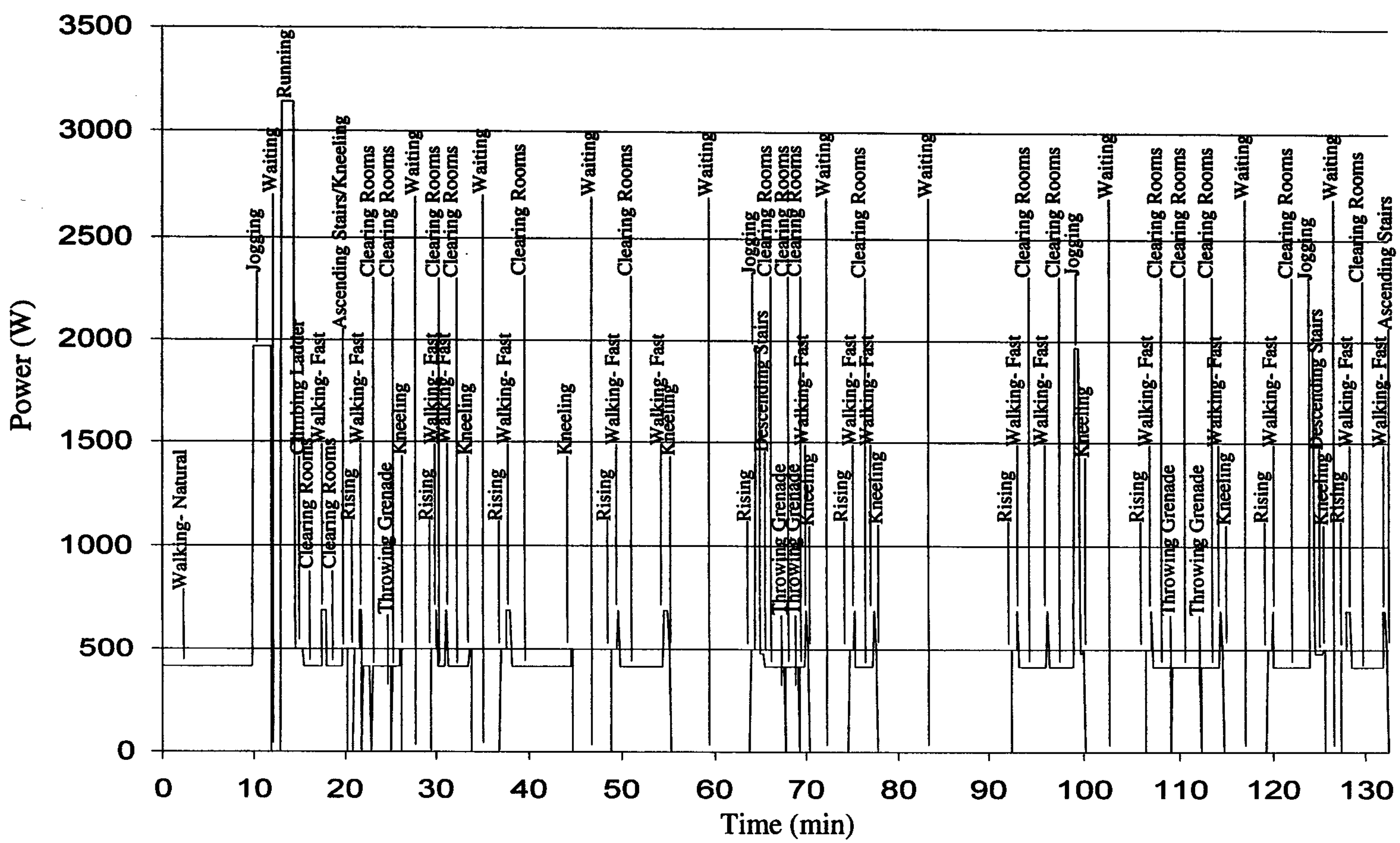

Figure B-3. Lower limb peak power profile for clear-building mission scenario. (Carried load throughout entire mission is fighting load.)

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13. ABSTRACT *(Maximum 200 words)*

This report provides baseline estimates of the power and torque requirements for the lower limbs of an exoskeleton[1] for two hypothetical dismounted missions. The missions are "movement to contact" and "clear a building". The power and torque estimates can be used to size key components such as the power supply and actuators. The estimates are based on human biomechanical data reported in journal articles and technical reports. The variability of biomechanical measures and the assumptions and potential errors involved in data collection and processing are discussed. The accuracy of the joint angle, torque, and power output variables is estimated to be within 20% of the reported values. In the use of these data, several assumptions are made regarding the equivalence of the exoskeleton and a human. To obtain peak power profiles for the hypothetical missions, each mission was broken into the tasks (walking, jogging, kneeling, jumping, etc.) that the soldier (50th percentile, by body mass, male carrying a load) performed. Then, the total power requirement over one cycle of each task was calculated, and the peak and average powers required during that cycle were identified. Combining the peak power requirements of the tasks in a chronological sequence results in a peak power profile for the entire mission. For the movement-to-contact mission, a nearly constant peak power of 500 W is required, with short-term requirements as great as 5.5 kW. For the clear-building mission, a nearly constant peak power of 400 to 700 W is required, with short-term requirements of 2 to 3.5 kW.

[1]A user-worn device that augments and enhances the wearer's speed, strength, and endurance

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